THE UNIVERSITY
of EDINBURGH

This thesis has been submitted in fulfilment of the requirements for a postgraduate degree (e.g. PhD, MPhil, DClinPsychol) at the University of Edinburgh. Please note the following terms and conditions of use:

This work is protected by copyright and other intellectual property rights, which are retained by the thesis author, unless otherwise stated.
A copy can be downloaded for personal non-commercial research or study, without prior permission or charge.
This thesis cannot be reproduced or quoted extensively from without first obtaining permission in writing from the author.
The content must not be changed in any way or sold commercially in any format or medium without the formal permission of the author.
When referring to this work, full bibliographic details including the author, title, awarding institution and date of the thesis must be given.

Whole Body Vibration: Stimulus Characteristics and Acute Neuromuscular Responses

by

Mark F. Sanderson

THE UNIVERSITY *of* EDINBURGH

A Thesis Presented for the Degree of Doctor of Philosophy

in

The Institute for Sport, Physical Education and Health Sciences
The University of Edinburgh

2014

Abstract

Whole body vibration (WBV) delivers a stimulus to the body via an oscillating platform and remains a relatively new area of research. Several applications of WBV stimuli have been developed as strength training and rehabilitation modalities, but inconsistent results have been published. There is little knowledge underpinning the mechanisms to explain the elicited neuromuscular responses to WBV and a wide range of WBV parameters across the literature. As a result, safe and effective protocols are yet to be established or validated. The aim of this current research was to investigate: the electromyography (EMG) and explosive performance responses to varying WBV frequencies; the effect of WBV data analysis techniques; and the influence of external factors on WBV stimulus and neuromuscular responses. Three main studies were completed:

1. An individualised response of both EMG and jump performance appears to exist dependent on vertical WBV frequency, in trained participants. This is in spite of no overall frequency dependent effect of EMG or performance responses across participants as a group. The influence of the role of expectancy effect appears minimal following this particular WBV protocol.
2. There was a significant effect of filter technique on EMG data recorded during vertical WBV. A tailored, WBV specific notch filter technique may offer an effective balance; excluding WBV noise artifacts without removing significant portions of valuable muscle signal EMG data.
3. The influence of external load on WBV acceleration output also appears minimal. Platform acceleration output was dependent on WBV frequency, as expected. Lower accelerations were recorded in superior body segments, suggesting a dampening mechanism, which was also proportionally dependent on frequency. EMG activity of upper and lower leg segments may differ in response to frequency, likely due to transmission distances involved. This may partially account for a potential dampening mechanism.

In addition, a protocol to quantify WBV stimuli delivered by this particular WBV type illustrated significant differences in theoretical and actual parameters. This may explain not only the lack of overall explosive performance effect reported earlier; but also the inconsistent WBV literature. Future research should quantify WBV stimulus before investigating possible neuromuscular responses to individualised protocols, which may be assessed via EMG activity.

Contents

Acknowledgments

My first acknowledgement is to Dr Tony Turner, who has provided such invaluable expertise and knowledge. Tony's approach to the role of principle supervisor could not have been better and I will be forever grateful for his encouragement and support. My second acknowledgement is to Dr John Sproule, whose vast experience of the PhD process was extremely useful and meant I was fortunate enough to have an ideal supervisory team, making these last five years very enjoyable. Both Tony and John have been very approachable, providing constructive feedback and only asking me to be a running partner and occasional computer guru in return.

My thanks go to Dr David Saunders who provided excellent advice regarding study design, also to Dr Simon Coleman who provided fine advice on data analysis. Everyone at Edinburgh University, Institute of Sport, Physical Education & Health Sciences made these five years so enjoyable. A special mention to Dr Jon Kelly (senior technician), who provided such thorough equipment support and time, even though I was one of "those PhD students". His teaching topics including wrapping up electrical wires correctly were informative. To Jon's able assistant, Maria, who provided often welcome entertainment and distraction, I thank you.

Many thanks to Dr Shaun Phillips, who offered advice and encouragement before setting an example by trailblazing a path to employment and the real world. To Edward Hall for becoming a short term flatmate but a long term friend; and to all in the PhD group, illustrating there is strength in numbers.

To my younger brother Ross, thank you for keeping track of all the times I announced a self-imposed final completion deadline, but more importantly for offering to meet for a suitable beverage. To my youngest brother Chris, thank you for keeping track of which letters (P. h. D.) I had achieved at appropriate milestones. My final acknowledgement is reserved for my parents, who have provided me with unwavering love and support. You have helped me make this possible and I hope to "repay" you somehow. Finding myself in permanent employment is a start!

Declaration

I hereby state that the following thesis has been composed by the student, Mark F. Sanderson and that the work is the student's own.

I declare that the work presented here has not been submitted for any other degree or professional qualification except as specified.

Mark F. Sanderson

Signed…………………………………………………………………………………….

Date……………………………………………………………………………………….

Abbreviations

°	degrees
°/s	degrees per second
% CV	percentage coefficient of variation
Δ0 Hz	pre-post change for 0 Hz
Δbest	best pre-post change
↓	decrease of significant level ($p < 0.05$)
↑	increase of significant level ($p < 0.05$)
A/D	analogue to digital
AF	average force
Ag	silver
ANOVA	analysis of variance
AP	average power
A_R	resultant acceleration
AV	average velocity
BM	body mass
Ca^{2+}	calcium
CJ	continuous jump
cm	centimetre
$cm.s^{-2}$	centimetre per second
CMJ	countermovement jump
CMRR	common mode rejection ratio
Cont RJ	continuous rebound jump
CSA	cross-sectional area
CT	contact time
dB	decibels
df	degrees of freedom
DF	dorsi-flexion
DJ	drop jump
D_{PTP}	peak-to-peak displacement
EMD	electromechanical delay
EMG	electromyography
EMG_{rms}	electromyography root mean square
eVDV	estimated vibration dose value
exs	exercise
FFT	Fast Fourier Transformation
fit	fitness group
f_{time}	flight time
g-forces	gravitational forces
GTO	golgi tendon organ
HR	heart rate
H-reflex	Hoffman-reflex
HR_{max}	maximum heart rate
Hx	history
Hz	hertz
ICC	intra-class coefficient
IMVC	isometric maximal voluntary contraction
ISO	international organization for standardization

iWBV	individualised whole body vibration
JH	jump height
kg	kilogram
m	metre
$m.s^{-2}$	metres per second squared (acceleration)
mV	millivolts
MFGC	maximum force generating capacity
min	minute
mm	millimetre
MRFR	maximal rate of force rise
MRI	magnetic resonance imagining
ms	millisecond
MVC	maximal voluntary contraction
M-wave	motor response
nEMG	normalised electromyography
N	newton
$N.s^{-2}$	newtons per second squared
NS	non-significant
PAD	post activation depression
PAP	post activation potentiation
PF	plantar-flexion
QL	quadratus laborum
r^2	coefficient of determination
RAMM	relative apparent mass magnitude
rec	recovery
res	resistance
RFD	rate of force development
RM	one repetition maximum
RSI	reactive strength index
s	second
SMRFR	stimulate maximum rate of force rise
SD	standard deviation
SQJ	squat jump
SSC	stretch-shortening cycle
TMS	transcranial magnetic stimulation
TVR	tonic vibration reflex
vel	velocity
VGRF	vertical ground reaction force
VMRFR	voluntary maximum rate of force rise
VT	vibration training
WBV	whole body vibration
α	alpha
γ	gamma
µm	micrometre

List of Figures **Page**

List of Tables **Page**

Chapter 5

Chapter 7

Chapter 1: Introduction

1.1 BACKGROUND

It has been previously established that improvements in neuromuscular performance can be due to adaptations from training relating to a specific stimuli, e.g. resistance training (Moritani & deVries, 1979). These strength training adaptations can improve neuromuscular performance via two widely accepted mechanisms. Firstly, via early neural adaptations (Sale, 1988) and secondly via longer term myogenic adaptations (McDonagh & Davies, 1984); often involving chronic training programmes (Bosco et al., 1999a). Both adaptations allow for greater force generating capacities in individuals, expressed as performance gains.

These adaptations stem from specific training stimuli, which can be the individual's own body weight or an external load (as in resistance training). As training methods have developed other external stimuli have been considered, one of which is vibration.

The exposure of the human body to vibration has received substantial scientific investigation, see Rittweger, (2010). The majority of this has involved identifying the potentially harmful effects of vibration. Such research has provided a well accepted consensus that chronic occupational exposure to vibration, can have several negative effects such as intervertebral disc displacement, hearing loss or vestibular damage (Griffin, 1990). For further information relating to International Organization for Standardization (ISO) Standards refer to Chapter 2, section 2.6. Vibration training (VT) has the potential for harmful effects (Jordan *et al.*, 2005; Prisby *et al.*, 2008; Stewart *et al.*, 2009; Wilcock *et al.*, 2009) and inappropriate vibration protocols (in terms of the magnitude of vibration parameters) may lead to increased injury risk (Lorenzen et al., 2009). However, it is important at this point to clarify that use of VT as a training stimulus has very different parameters to that of occupational vibration (which commonly involves higher magnitudes of stimuli and more prolonged exposures).

Over the last 15 years the use of VT has grown in popularity, both in research and in commercial use. This research has involved: low frequency (5 – 60 Hz); low peak to peak displacement (2 – 10 mm); and often relatively short exposure time (30 – 90 s) (Rittweger, 2010). The research has involved locally applied vibration in which the vibration stimulus is directly applied often to a muscle tendon (Luo et al., 2005b); or indirect exposure to vibration via a specifically manufactured device. Most commonly this is a platform on which an individual stands or a vibrating dumbbell for upper body exposure (Bosco *et al.*, 1998; Bosco *et al.*, 1999a).

Initially, research into the effectiveness of VT investigated vibration as a stimulus to enhance neuromuscular mechanisms during strength and power performances. One of the main reasons for such interest stemmed from work by Eklund & Hagbarth (1966) with the identification of the tonic vibration reflex (TVR). Work by Matthews (1966) and Burke *et al.* (1976b) proposing the involvement of a muscle spindle driven stretch reflex, furthering the scientific interest regarding VT. In the 1990's research interest in VT grew due to strength and power gains following acute exposure to vibration (Issurin *et al.*, 1994; Bosco *et al.*, 1999a). But also following exposure to whole body vibration (WBV) (Bosco et al., 1998). Possible mechanisms responsible for such increases in neuromuscular performance were proposed, such as modulation of muscles stiffness (Cardinale & Bosco, 2003), and several others have been proposed since; see review paper (Rittweger, 2010). These will be discussed in Chapter 2. As with most new training stimuli it generated good scientific debate and conflicting research findings, for example: Cochrane & Stannard (2005) *cf.* Bullock *et al.* (2008); as well as conflicting reviews, for example: Rehn *et al.* (2007) *cf.* Nordlund & Thorstensson (2007).

More recently there have been a number of studies investigating the wider application of vibration, both for other performance indicators, such as flexibility, for example: Kinser *et al.* (2008); McNeal and Sands, (2006); and Sands *et al.* (2006) but also as a therapeutic tool, for example: Trans *et al.* (2009). The research areas of VT with regards flexibility and proprioception have increased in terms of the number of publications (Rittweger, 2010). The application of vibration as a therapeutic tool, specifically post-injury, has only recently received limited attention thus far (Moezy et al., 2008). Additional areas which have been explored in the literature include (but not exclusively) the effect of vibration on: hormones (Erskine et al., 2007); circulatory effects (Lythgo et al., 2009); elderly populations' balance, mobility and osteoporotic status (Furness *et al.*, 2010; Lau *et al.*, 2011); and finally on neurological clinical populations (Sitja Rabert et al., 2012). All of the fore-mentioned areas of vibration are important and continued research is warranted. However, these areas are outside the remit of this thesis, and thus will not be covered.

As the research interest in VT expanded, so too did the commercial applications available for sport teams and public gyms. This has been most common via a WBV platform, of which there are several manufacturers. With growing popularity and availability of such devices it increases the need for research-based protocols, which are both effective but more importantly, safe. However, what is apparent from the literature is the lack of proven VT protocols in terms of: effective intensity and exposure time of safe vibration stimuli (Cardinale & Wakeling, 2005; Adams *et al.*, 2009); increases in acute neuromuscular explosive performance; as well as proposed mechanisms. This will be expanded upon in later sections. In summary, the wide variety of VT parameters and inconsistencies in protocols make prescriptions of ideal vibration stimuli difficult (Rehn *et al.*, 2007; Bazett-Jones *et al.*, 2008a). Lack of research into the effectiveness of different vibration parameters highlights the lack of rationale as to why one setting should be used over another (Cardinale & Lim, 2003b; Jordan *et al.*, 2005). Standards for the use of vibration as a training stimulus are lacking and tend to be anecdotal (Prisby *et al.*, 2008; Stewart *et al.*, 2009). Underpinning the literature is debate regarding the mechanisms responsible for WBV responses. Overall there appears a lack of consensus across both general and athletic populations.

1.2 PURPOSE OF THE THESIS

The initial aim of this thesis was to investigate the influence of changing WBV parameters (namely frequency) and what possible effect that may have on acute performance responses. The aim was to further emerging research which alluded to a possible individualised WBV response as identified in Chapter 2 literature review. Chapter 4 represents an opportunity which arose to investigate this in elite athletes and in a sense formed an initial study. Whereas, Chapter 5 represents further investigation into this area but in a more controlled and standardised lab setting. Following on from these findings a mechanistic approach was taken in Chapter 6 to investigate the influence of data analysis methods (in this case electromyography (EMG) filtering techniques) which may have accounted for the results of Chapter 5. As a consequence of the findings presented in Chapter 6, the purpose of Chapter 7 was to investigate a second potential reason which may account for the original findings; the influence of external load on WBV stimuli output. This chapter also investigated the characteristics of other WBV parameters delivered by this particular WBV platform. Finally, Chapter 8 concludes with a general discussion which provides a summary of the: acute neuromuscular responses; data analysis techniques; and the quantifying of WBV magnitudes. This is with an over-arching aim to provide rationale for future WBV protocols.

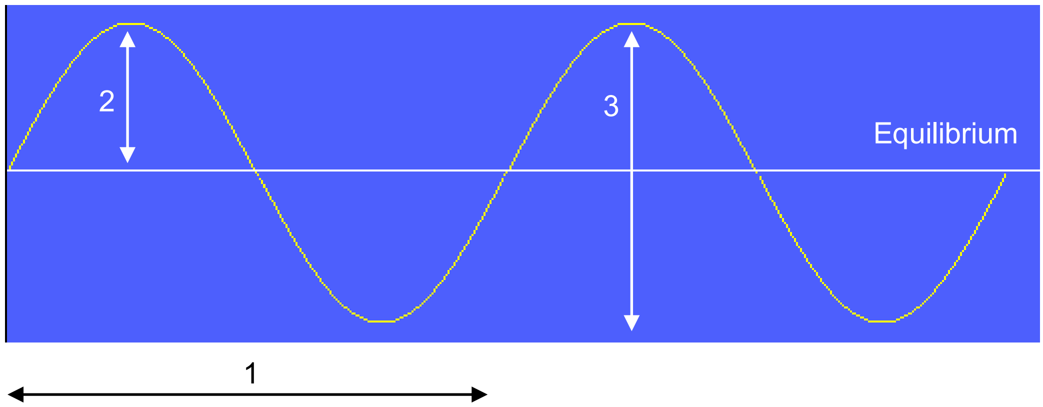
Therefore, the overall aim of this thesis was to characterise WBV stimulus and the acute neuromuscular responses of such stimuli, in an attempt to inform the rationale behind safe and effective WBV protocols.

Chapter 2: Literature review

2.1 VIBRATION

Vibration can be defined as an oscillatory motion (Griffin, 1990; Cardinale & Wakeling, 2005). Vibration is not an uncommon phenomenon; in fact the body is exposed to such a stimulus regularly through activities such as walking and running (Wakeling et al., 2002). Various forms of vibration exist, however sinusoidal vibration allows investigation into the response of such a stimulus (Griffin, 1990). The frequency of vibration is characterised by the number of oscillation cycles repeated in a set period of time, most commonly per second giving rise to the unit hertz (Hz), Figure 2.1.

Figure 2.1: Sinusoidal oscillatory vibration motion. 1, oscillation frequency (Hz); 2, oscillation amplitude (mm); 3, oscillation peak to peak displacement (mm).



A unit of vibration is ‘amplitude’ which describes the magnitude of the oscillation displacement. This unit is often described as the peak displacement, peak to peak displacement (D_{PTP}) or acceleration. Throughout the literature terms such as amplitude and D_{PTP} have been interchangeable, highlighting inconsistency describing vibration characteristics, Figure 2.1 (Lorenzen et al., 2009). For clarification, amplitude is defined as the maximum displacement of an oscillating body from its equilibrium position (Lorenzen et al., 2009). The D_{PTP} better describes the total magnitude of oscillation between the positive and negative extremes (i.e. D_{PTP} is amplitude multiplied by 2, Figure 2.1) (Lorenzen et al., 2009).

Therefore the use of D_{PTP} has been advocated (Cardinale & Bosco, 2003; Lorenzen *et al.*, 2009; Rauch *et al.*, 2010) as it allows further calculations to determine the gravitational acceleration (g-force) of the vibration, based on the Earth's gravitational acceleration (g = a constant of 9.806 $m.s^{-2}$) (Wilcock et al., 2009):

$$\text{g-force} = \frac{D_{PTP}\,(2\pi f)^2}{2g\sqrt{2}} \qquad \text{(Equation 2.1).}$$

g-force = gravitational acceleration.
D_{PTP} = peak to peak displacement in metres (m) not millimetres (mm).
f = frequency; the rate of an oscillation cycle.
g = gravity constant = 9.806 $m.s^{-2}$.

(Wilcock et al., 2009).

g-force can be expressed as acceleration (acc):
acc ($m.s^{-2}$) = g-force x 9.806 $m.s^{-2}$ (Equation 2.2).

(Rittweger, 2010).

The advantage of calculating the gravitational acceleration of a vibration oscillation is that allows comparisons between vibration stimuli across the literature. As demonstrated there are several parameters associated with both VT and WBV introducing many connotations to training protocols. This is not to mention the differences in delivery methods of vibration stimuli.

2.2 DIFFERENCES IN ACUTE VIBRATION PROTOCOLS

2.2.1 Locally applied vibration versus whole body vibration

Locally applied vibration differs from WBV as the vibration stimulus is applied directly to the muscle or tendon, via a portable vibration device, onto the skin;

whereas WBV delivers the vibration stimulus via an oscillating platform in which the individual stands upon. There appears a lack of consensus on the effect of locally applied vibration on performance measures. For example, prolonged exposure (30 minutes) has been shown to significantly reduce power output (Jackson & Turner, 2003); compared to 10 minutes which increased muscle torque mechanisms (Brunetti et al., 2006).

Recommendations for optimal parameters of locally applied vibration, in terms of elicited muscle activity during vibration, have been made by Luo *et al.* (2005b) (65 Hz frequency and 1.2 mm D_{PTP}). However, these were based on upper body exposure to vibration, locally applied to the bicep tendon. Further conflict exists as subsequent research found no influence on velocity or power parameters (Moran *et al.*, 2007; Luo *et al.*, 2009). Reasons for the conflicting findings may be that different vibration stimuli are required to optimise the response during maximal performance (Moran et al., 2007).

For vibration applied locally to the lower body, there have been positive physiological effects reported such as increased quadriceps strength and EMG activity (Mileva et al., 2006) and attenuation of strength decreases during muscle soreness (Bakhtiary et al., 2007). Although work by Mileva *et al.* (2006) involved a vibration like stimulus superimposed during knee extension, applied through machine weight apparatus; whereas vibration was directly applied to quadriceps, hamstring and calf muscle bellies by Bakhtiary *et al.* (2007). But conflict still exists regarding lower body responses to vibration. No change in velocity or power parameters following direct vibration of the quadriceps muscle tendon was observed (Luo et al., 2008). Luo *et al.* (2009) suggested that direct vibration may not be able to enhance neuromuscular output due to the direction the vibration stimulus is applied, (perpendicular to the muscle length). Indirect vibration (i.e. WBV) may cause small but rapid changes in muscle length parallel to the length of the muscle (Cardinale & Bosco, 2003). Whereas, locally applied vibration may result in a more perpendicular direction of stimulus, affecting any potential neuromuscular enhancement (Luo et al., 2009). Furthermore, the recommended parameter (65 Hz, 1.2 mm) may not be appropriate for the lower body as this was measured with respect to the upper body (Luo et al., 2008).

Differences between locally applied vibration and WBV include a dampening effect of the vibration magnitude by soft tissues during WBV. This may mean that the stimulus parameters (such as D_{PTP}) reaching the target muscles may be very small (Luo et al., 2005b). This can be influenced by the distance from transmission source (the WBV platform) and the targeted muscles (Abercromby et al., 2007b). The actual WBV parameters that are exposed to the target muscles may be difficult to quantify, whereas locally applied vibration experiences a reduced transmission distance and thus dampening effect (Brunetti et al., 2006).

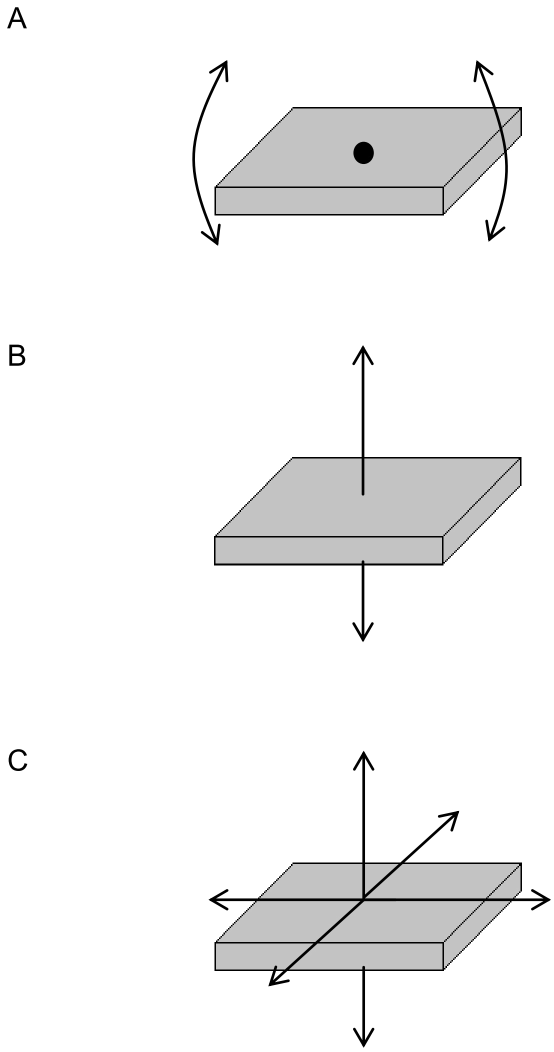
Clearly there is a difference in VT protocols between locally applied vibration and WBV. This literature review is focused on WBV although, even within WBV there are significant differences evident.

2.2.2 Different types of whole body vibration

As the amount of research regarding WBV training has increased over the last 10 years several commercial manufacturers have produced WBV platform devices. There are numerous platforms available e.g.: Galileo™ (Novotec), NEMES™ (Bosco-System); VM Master (VibraMachine); and Powerplate® (Powerplate Ltd). These WBV platforms differ in appearance but more importantly in the type of WBV delivered. WBV platforms can be characterised by the direction of oscillation, Figure 2.2. Galileo platforms provide a rotational oscillation output around a pivot. The NEMES and VM Master platforms provide a vertical oscillatory output, whereas Powerplate platforms provide oscillation in a majority vertical direction but also in a horizontal direction (known as tri-planar) (Powerplate, 2013). The physiological effects of these differing WBV mechanisms is largely unknown as very few studies have directly investigated this aspect (Wilcock et al., 2009). This makes comparisons across the literature very difficult which will be discussed in sections

2.4, e.g. Adams *et al.* (2009) *cf.* Bullock *et al.* (2008) with regards a WBV response in counter-movement jump (CMJ) performance.

Figure 2.2: Whole body vibration platform types and direction of oscillation output. A, Galileo (rotational); B, NEMES and VM Master (vertical); C, Powerplate (tri-planar).



Abercromby *et al.* (2007a) directly compared Galileo WBV and Powerplate WBV reporting significantly higher EMG activity in vastus lateralis (103 versus 77 % above baseline) and gastrocnemius (151 versus 132 % above baseline) during Galileo versus Powerplate WBV respectively, suggesting different WBV types may elicit different neuromuscular responses. Different magnitudes of vibration stimulus transmission were reported; 71 to 189 % greater transmission during vertical versus rotational WBV (Abercromby et al., 2007b). Limiting these findings was pre exposure differences in EMG activity between groups, suggesting inherent differences before WBV commenced. However, Ritzmann *et al.* (2012) supported the findings by Abercromby *et al.* (2007a) reporting greater EMG activity (66 to 91 %) during Galileo versus Powerplate WBV platforms. Again transmission characteristics were suggested to influence WBV platform specific EMG responses (Ritzmann et al., 2012). Although, the possibility of WBV induced motion artifacts influencing the magnitude of EMG response should be highlighted, which will be discussed in chapter 6.

Further limiting the findings of both Abercromby and co-workers and Ritzmann and co-workers were: short WBV exposure times (15 and 10 s respectively); and utilising only 30 Hz WBV frequencies. The aim of the two studies did not address strength or power measures in relation to WBV platform type. One study which did utilise a performance measure found no difference in CMJ force or force development between three WBV platform types (Bagheri et al., 2012). However, WBV exposure times were short (15 or 40 s) and jump performance measures were averaged across three attempts. Marin & Rhea (2010b) suggested that for chronic WBV (repeated over a period of at least one week), vertical WBV was more effective than rotational platform type. Yet the meta-analysis review authors admitted there were moderating factors: rotational WBV platforms are unable to generate higher output frequencies than 30 Hz; rotational WBV training exposure times tended to be lower than vertical WBV programmes (e.g. 267 versus 464 s); chronic programmes involving rotational WBV platform types tended to be shorter in exposure time than vertical WBV programmes (e.g. 10 weeks versus 13 weeks). Therefore, further research is warranted on the acute performance responses following different WBV types.

2.2.3 Whole body vibration parameters

During exposure to WBV there are several parameters (see section 2.1) that can be adjusted (frequency, D_{PTP}, exposure time and posture) and therefore numerous combinations of WBV protocols. There is a lack of consistency in the literature regarding WBV parameters and uniform terminology (Lorenzen et al., 2009). Optimal combinations of WBV parameters remain unknown (Da Silva-Grigoletto et al., 2011). With the potential for injury if inappropriate parameters are used (e.g. hand-arm vibration syndrome and lower back pain (Jordan et al., 2005), see section 2.6); researchers must be able to clearly reproduce and justify WBV protocol.

2.2.3.1 Frequency

Frequency appears the most consistently reported (Lorenzen et al., 2009) and has had the most research completed investigating the influence of changing this parameter. Despite this, knowledge of safe, effective and appropriate VT protocols in relation to frequency settings is limited (Cardinale & Wakeling, 2005). Further work is required to determine the optimal WBV protocols (Prisby et al., 2008) and the variety of protocols (10 to 90 Hz frequencies; 0.1 to 6 mm D_{PTP}) may explain the inconsistent evidence across the literature (Kiiski et al., 2008).

EMG activity responses to WBV may be frequency dependent (Fratini et al., 2009a). The EMG activity for triceps surae was approximately 40 % more dependent on frequency than quadriceps based on Galileo WBV frequencies of 5 to 30 Hz (Ritzmann et al., 2012). Varying WBV training protocols (such as WBV parameters) has the potential to have a large effect on EMG parameters of lower leg muscle groups. Recently there has been an emerging argument that the response to WBV may well be individual specific (Cardinale & Lim, 2003b).

Di Giminiani *et al.* (2009) proposed that each individual has an optimal frequency of WBV at which the greatest response may be elicited (see chapter 5). WBV at a frequency similar to the natural resonant frequency may elicit the highest muscle activation (Cardinale & Wakeling, 2005). The natural resonant frequency of soft tissue for example is the frequency content at which a shock wave produced by a repetitive impact force travels through said soft tissue (e.g. lower leg musculature during heel strike) (Nigg & Wakeling, 2001). However, there is debate regarding which frequency would most efficiently elicit the TVR (Savelberg et al., 2007). A review (Wilcock et al., 2009) suggested that 30 Hz may be an optimal frequency for eliciting the highest EMG activity response. The authors acknowledged a possible individual response to WBV training. The review stated that EMG analysis of individual athletes' response to varying WBV frequencies may be required to monitor the response. One such study (Cardinale & Lim, 2003b) found that in professional volleyball players 30 Hz elicited a significantly higher (34 %) quadriceps EMG response. Supporting this, CMJ power was significantly higher following 30 Hz versus 35 and 40 Hz, but not significantly different versus 50 Hz (Bedient et al., 2009).

Conflicting this was Cardinale & Lim, (2003a) who found enhanced squat jump (SQJ) performance by 4 %, citing that 20 Hz is similar to the resonant frequency of running and may explain the adaptive responses. The type of jump and therefore the jump biomechanics may affect the response that frequency could elicit with regard to differences in stretch shortening cycles (Di Giminiani et al., 2009). The effect of frequency may influence muscle fatigue and the TVR response, by either failing to induce muscle fatigue and triggering only a limited TVR response; or by eliciting a strong TVR response (Cardinale & Lim, 2003a) (see section 2.3.2). Although the work by Cardinale & Lim (2003a) is limited by a lack of volume-matched controls and it is also unclear if the study design was blinded to participants, raising the issue of a role of expectancy effect. The finding that 20 Hz enhanced SQJ performance is supported (Da Silva-Grigoletto et al., 2006). Each of the frequencies (20, 30, 40 Hz) investigated improved one of the performance measures such as SQJ, CMJ and 1 repetition maximum (RM). No one single frequency increased all outcome measures greater than other frequencies (Da Silva-Grigoletto et al., 2006). No EMG data was collected so it is unclear what the effect of these frequencies on neuromuscular properties would have been.

Di Giminiani *et al.* (2009) did record EMG activity during different WBV frequencies. A chronic individualised frequency (the frequency eliciting the highest EMG response) WBV programme (8 weeks) elicited an 11 % increase in squat performance versus a 3 % increase following a chronic standardised WBV programme. Therefore, in terms of frequency, an individualised approach may result in more effective response to WBV training. If individualised frequencies for WBV programmes maximise the response to WBV this may explain the conflicting findings across the literature, which have utilised standardised frequencies for all participants. Some of these WBV frequencies will not have been optimal frequency for the participants.

Further conflict exists with regards optimal WBV frequency, as the influence of frequency could be gender dependent. Both 40 and 50 Hz significantly increased female but not male CMJ performance (Bazett-Jones et al., 2008a). However, the study design simultaneously changed other WBV parameters such as D_{PTP}, limiting the conclusions drawn. In addition, the displacement data during the CMJ performance was obtained via a linear encoder attached to an Olympic bar held at shoulder height. This would be subject to movement which would not have been standardised. It is clear that the influence of WBV frequency is conflicting: (recommended frequencies of 20 and 30 Hz *cf.* frequency does not affect performance). Different types of outcome measures ranging from single jointed isometric to multi jointed dynamic tests may influence any conclusions (Savelberg et al., 2007).

Squat jump performance in female only participants was not affected by frequency (Gerodimos et al., 2010). However, this was based on 6 minutes of continuous Galileo WBV and therefore could only involve frequencies of up to 30 Hz. The isometric posture of 170 º knee angle adopted during WBV limits the use of a SQJ performance commencing from 90 º knee angle as an outcome measure.

A stronger methodological study (Adams et al., 2009) recruited participants of both sexes and investigated the effect of different frequencies of WBV on a dynamic, multi-jointed CMJ performance which is applicable to athletic performance. A combination of low frequency with low D_{PTP} (30 Hz, 2-4 mm) and high frequency with high D_{PTP} (50 Hz, 4-6 mm) elicited increased CMJ power performance (by 1 and 1.6 % respectively). Conflicting this finding, Armstrong *et al.* (2010) reported no frequency effect on CMJ performance utilising an identical WBV platform type. Reasons for this include mixed gender groups and different techniques assessing CMJ performance (vertical jump tester *cf.* a criterion method, e.g. force platform). The vertical jump tester relies on jump flight time to determine the change in height of the centre of gravity and was calibrated to approximately 1.3 cm. To complete the vertical jump test upper limb involvement is required, which is known to influence jump performance (Lees et al., 2004). This is in comparison to research which utilises ground reaction forces obtained from an accurately calibrated force platform involving a standardised hands-on-hips protocol. A meta-analysis (Marin & Rhea, 2010a) suggested there was only a trend in chronic (at least one week repetition period) WBV towards a dose response to frequencies of less than 35 Hz. The relatively unknown effect of frequency on performance and the wide variety of WBV frequencies used may explain the contradictory findings across the literature (Rehn et al., 2007).

2.2.3.2 Peak to peak displacement

As with frequency, D_{PTP} has varied through the literature from 3.4 to 10 mm (Rehn et al., 2007). There are few studies that have solely investigated the effect of varying D_{PTP} on performance. Adams *et al.* (2009) investigated varying the D_{PTP} in conjunction with frequency, quoting a range of D_{PTP}. The significance of a D_{PTP} range is that the g-forces exposed to the participants will also have a range (see equation 2.1). This is relevant as comparing WBV exposures expressed as ranges may be problematic. The data analysis by Adams *et al.* (2009) appears to have focussed on frequency even though a time x D_{PTP} interaction was identified as significant, with high D_{PTP} producing greater CMJ power values post WBV than low D_{PTP} (at 40 and 50 Hz, CMJ power increased 1 – 1.6 % versus 0.3 – 0.5 % following high and low D_{PTP} respectively).

Bedient *et al.* (2009) suggested D_{PTP} had minimal influence on CMJ power performance as 30 Hz WBV produced higher CMJ power versus 35 and 40 Hz regardless of the D_{PTP}. Supporting this; Bazett-Jones *et al.* (2008a) measured acceleration while varying D_{PTP} (2 – 4 mm versus 4 – 6 mm) in a blinded study design. In male participants varying peak to peak displacement did not have an effect on CMJ performance. However, in female participants CMJ performance was significantly improved (by 9 % following 2 – 4 mm D_{PTP}; and by 8% following 4 – 6 mm D_{PTP}). This suggests a gender dependent response as with frequency. However, as both frequency and D_{PTP} can determine acceleration and were changed simultaneously it is unknown which may account for the enhancement of CMJ performance. Armstrong *et al.* (2010) supports Bedient and co-workers as no effect of D_{PTP} was found on CMJ performance. Gerodimos *et al.* (2010) reported no effect of D_{PTP}, albeit on SQJ performance and during 25 Hz only.

Further study designs should recreate similar g-forces and keep one variable constant. The ability to quantify acceleration is important, as g-forces may be different across platforms (see chapter 7). The full effects of altering D_{PTP} on performance are unclear at present.

2.2.3.3 Posture

Participant's posture and therefore the length of the muscle during WBV exposure is an important aspect as muscle length may determine TVR response (Nordin & Hagbarth, 1996). Therefore, authors have proposed that WBV stimulation may be more effective when applied to stretched muscles (Cardinale & Lim, 2003b; Ritzmann *et al.*, 2012). For example, a half squat posture adopted on a WBV platform is an effective posture to ensure the quadriceps are in a stretched state (Cardinale & Lim, 2003b). However, the conclusions by Nordin & Hagbarth (1996) were based on in vitro studies with locally applied vibration and not WBV. The work by Cardinale & Lim, (2003b) did involve WBV although the investigation into effect of joint angle and thus muscle length on EMG activity during WBV was not the aim of the study. Posture appears to influence energy absorption and vibration stimulus transmission, although this work was only based on 25 Hz (Berschin & Sommer, 2010).

Higher knee flexion angles increased EMG activity in the quadriceps muscle group but decreased triceps surae EMG activity (Ritzmann et al., 2012). This was based on knees flexion angles of only 10, 30 and 60 º and is limited by not investigating one of the common postures adopted during WBV of 90 º knee flexion.

Abercromby *et al.* (2007b) investigated the effect of knee angle on relative apparent mass magnitude (RAMM) and head acceleration. The RAMM can be equated to the body's impedance, a higher RAMM is associated with an increased absorption of vibration energy (Abercromby et al., 2007b). Knee angle inversely affected RAMM and the smallest knee angle (10 – 15 º) gave significantly higher RAMM and head accelerations (Abercromby et al., 2007b). The authors concluded that the greatest amount of mechanical energy transmission to the upper body and head occurred at low knee angles and these should be avoided. A number of limitations with the study should be highlighted. The study made recommendations based on ISO standards, which were calculated for prolonged vibration exposure (i.e. occupational) based on sitting postures not standing (Abercromby et al., 2007b). The study is also limited by only investigating knee angles between 10 – 35 º and not 45 – 90 º which are more commonly adopted in WBV research. Therefore the conclusions regarding posture cannot be generalised across WBV literature.

One such study (Roelants et al., 2006) investigated the EMG response of more commonly used knee angles (55 – 90 º) in quadriceps. No effect of knee angle on EMG activity was recorded, conflicting with the theory of a greater WBV response with stretched muscles. This may have been due to different WBV parameters than those studies which did report EMG responses to posture. Further research is required investigating the effect of posture on the response of WBV in both EMG and performance outcomes.

2.2.3.4 Exposure time

The general lack of knowledge of the efficiency of WBV training protocols is especially apparent with the parameter of time. To date there are few published investigations into the effect of time in WBV exposure. Presently there is no scientific evidence suggesting certain exposure times of WBV elicit a significantly greater effect than other exposure times. Although a meta-analysis suggested 12 to 15 minutes of WBV stimulus per training session would elicit maximal gains (effect size of 2.1) (Marin & Rhea, 2010b). For studies investigating the acute effects of WBV (defined for this thesis as research into the effect immediately, or within 1 to 2 hours, post exposure); observational and anecdotal evidence gathered by this literature review suggests the exposure time has ranged from 5 s to 4 minutes with varying recovery periods between them. Similar evidence suggests the most commonly used WBV exposure time has been 60 s however, specific rationale and justification across the literature is lacking. Regarding chronic WBV studies observation suggests exposure times of 30 to 120 s have been utilised.

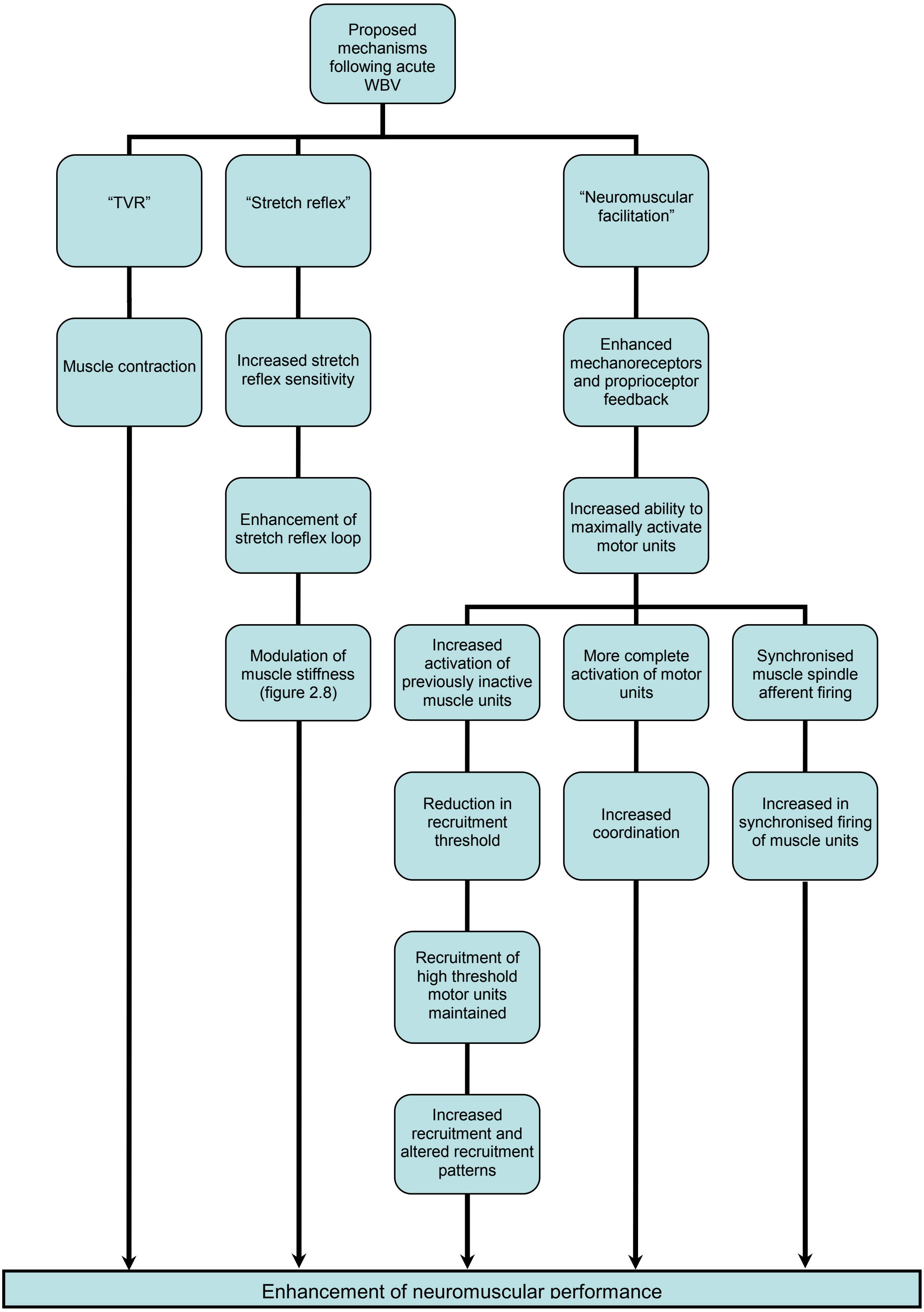
Adams *et al.* (2009) investigated the effect of exposure time amongst other variables of WBV. Exposure time of WBV (30, 45 and 60 s) had no significant effect on CMJ power. Stewart *et al.* (2009) found longer WBV exposure times of 4 – 6 minutes significantly reduced knee extension torque, whereas 2 minutes significantly increased torque. Although Stewart *et al.* (2009) utilised longer WBV exposure times than commonly used, the use of 5 º knee angle limits the conclusions as WBV responses may have been diminished due to lack of stretch applied to the quadriceps. Da Silva-Grigoletto *et al.* (2011) reported only 60 s exposure time increased SQJ, CMJ and loaded squat power performance, yet effect sizes were trivial (Cohen's *d* of 0.09 – 0.37). Ninety seconds WBV was reported to either decrease or have no effect on the same performance measures (Da Silva-Grigoletto et al., 2011).

Further work investigating repetition exposure time reported CMJ power and velocity was higher (mean differences of 23.78 Watts and 0.05 $m.s^{-2}$ respectively) following 3 x 10 s versus 1 x 30 s and may be more effective to increase jump power (Lamont et al., 2010a). However, this work is limited by the jump methodology involving taking a best pre WBV jump attempt and comparing it with an individual post WBV jump. Clearly the effect of WBV exposure time requires further research, (Marin & Rhea, 2010b) and the variation of stimulus exposure time across the literature makes interpretation of WBV efficiency problematic (Armstrong et al., 2010). This could also be said for periodisation of chronic WBV programmes, as only initial research has been completed (Dabbs et al., 2011); suggesting no overall significance difference in jump performance relating to the duration of rest interval. However, the same initial research suggested an individualised response to varying rest intervals may be evident.

2.3 WHOLE BODY VIBRATION AND PROPOSED MECHANISMS

The exact mechanisms that could account for potential responses to vibration are unclear and it has been argued that the mechanisms have not been fully explained (Cronin et al., 2004). Authors have admitted that little is known of the exact mechanisms (Armstrong et al., 2008) and a clear consensus is lacking (Rehn et al., 2007). There are a number of studies which have put forward proposed mechanisms either as general concepts (Adams et al., 2009) or proposed specific mechanisms, for example: TVR (Burke et al., 1976a); or muscle spindle reflex loop (Bosco et al., 1999a). The proposed mechanisms include: TVR (Eklund & Hagbarth, 1966); mechanoreceptor reflexive activity and neuromuscular facilitation (Cardinale & Bosco, 2003); hyper-gravitational forces (Nordlund & Thorstensson, 2007); muscle tuning (Cardinale & Wakeling, 2005); and post activation potentiation (PAP) (Cochrane et al., 2010), Figure 2.3.

Figure 2.3: Diagrammatic summary of proposed neuromuscular mechanisms responsible for increased performance following acute WBV.



Before these are discussed it would be useful to briefly introduce the anatomy and neurophysiology involved in the proposed mechanisms.

2.3.1 Anatomy and neuromuscular physiology

2.3.1.1 Mechanoreceptors

There are several different types of mechanoreceptors present within the musculotendinous unit. The most common is the muscle spindle (Zelena, 1994). Muscle spindles can range from 1 to 12 mm in length and 50 to 150 μm in diameter (Hunt, 1990). The muscle spindle is innervated by large primary afferent neurons, classified as Ia fibres. Gamma (γ) motor neurons play a role innervating certain types of fibres within the muscle spindle, but also serve to regulate the sensitivity of the muscle spindle. The fibres which receive innervations from Ia fibres form the primary sensory endings; and those received from group II afferent fibres form the secondary sensory ending. It is the primary sensory endings which respond to dynamic changes in muscle length (Hunt, 1990).

The overall function of muscle spindles is to provide sensory information with regards changes in muscle length. This occurs either by stretching the equatorial regions of the spindle by contracting polar regions resulting in local stretching of the spindle or; by passively stretching the equatorial regions caused by lengthening of muscle fibres. In both instances the muscle spindle is activated sending afferent impulses to the spinal cord. This forms a monosynaptic reflex loop with ά motor neurons supplying motor control innervations of the muscle. As a result of the muscle spindle stretch stimulus the muscle containing the spindles contracts via reflex, section 2.3.1.2 and Figure 2.4.

Another type of mechanoreceptor are Golgi tendon organs, which are spindle shaped receptors arranged around extrafusal muscle fibres located within the muscle tendon complex beyond the musculotendinous junction (Pocock & Richards, 1999). The tendon organ consists of two poles, a muscle pole which allows the receptor body to connect to a group of extrafusal muscle fibres, through individual tendons (Zelena, 1994). These mechanoreceptors monitor changes in muscle tension and are activated when the tendon organ is stretched by muscle contraction. Activation leads to neural signals being sent to the spinal cord via sensory neurons which, in association with inhibitory neurons, prevents excessive contraction of the muscle (Zelena, 1994).

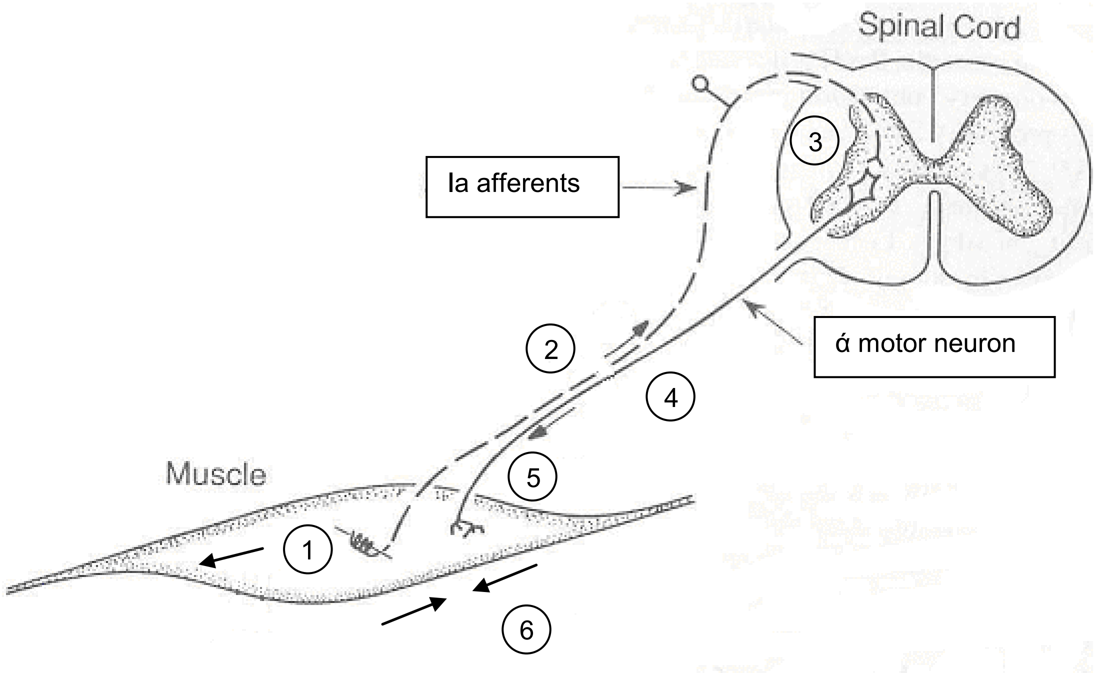
A third type of mechanoreceptor is a Meissner corpuscle, which monitors touch or flutter. These are rapidly adapting mechanoreceptors of the skin with low thresholds which respond to stimuli with a burst of discharge activity which ceases if the stimulus is maintained (Zelena, 1994). It would seem reasonable to suggest that the response to WBV from these mechanoreceptors is likely to be initially rapid then decline (Hamano et al., 1993).

2.3.1.2 Stretch reflexes

Stretch reflexes enable muscles to act in a spring-like manner in order to generate power (Enoka, 2002). Stretch reflexes can be characterised by the response to an unexpected increase in muscle length (Enoka, 2002). The neural circuit and processes involved are described in Figure 2.4. An example of a stretch reflex is the patellar tendon tap reflex. By striking the patellar tendon (causing a stretch) the quadriceps contract to minimise the rapid stretch of the tendon, therefore causing a knee jerk response (Enoka, 2002).

Figure 2.4: The stretch reflex (Enoka, 2002).

.



1 A stretch stimulus is detected by the muscle spindle.

2 As a response to the stretch, action potentials are generated from muscle spindles Ia afferents to spinal cord.

3 These synapse with ά motor neuron within the spinal cord.

4 The ά motor neuron innervates the muscle which was exposed to the stretch .

5 As the stretch stimulus is sufficient enough the ά motor neuron evokes a contraction within the muscle.

6 Overall the stretch stimulus elicits a contraction which aims to minimise the stretch.

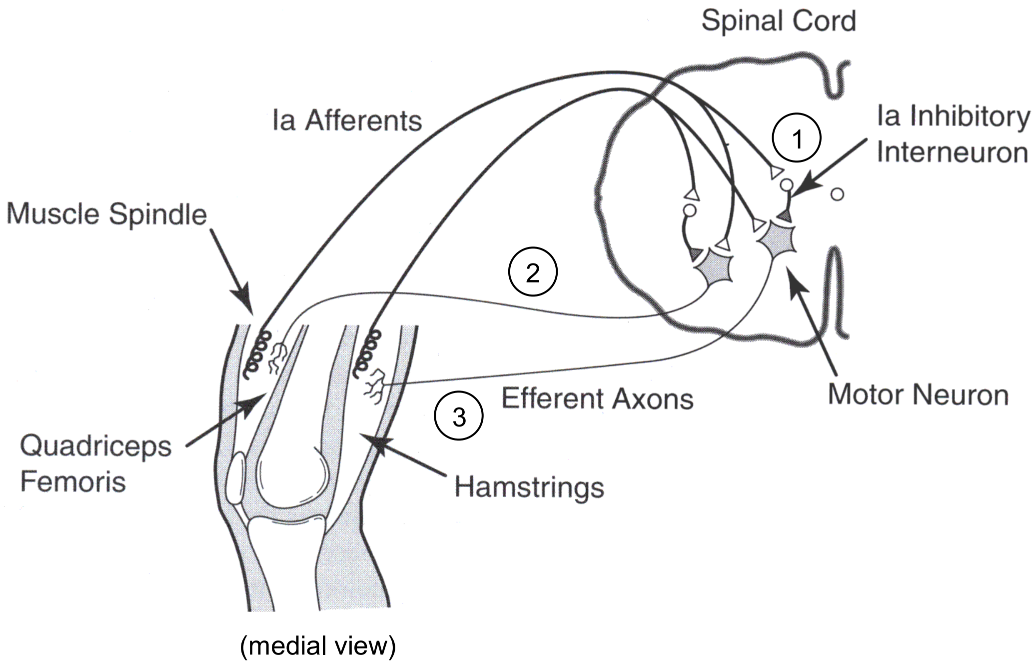
The stretch reflex response has been categorised into different components (Matthews, 1991) according to a time latency. Matthews (1991) has illustrated that the response to a stretch stimulus, as measured by EMG, begins soon after the stretch stimulus. The first category has been classified as a short-latency followed by a second category, the long-latency component (Matthews, 1991). However, there is some debate whether a third component is present; as it appears it is only occasionally observed (Enoka, 2002) or no longer widely recognised (Matthews, 1991). Similarly, there is debate over the time intervals of short and long latencies. Generally, these are at 30 and 60 milliseconds (ms) respectively (Enoka, 2002). However, conflicts exists with regards the time intervals of these latencies and it may vary between muscles (Matthews, 1991). Within the short latency component it is generally accepted that the stretch reflex activates both group Ia and II afferents via muscle spindles (Enoka, 2002).

2.3.1.3 Reciprocal-Inhibition reflex

There are other reflexes associated with the stretch reflex and these involve muscle spindle Ia afferents branching within the spinal cord. One of which is with a Ia inhibitory interneuron (Enoka, 2002). These interneurons have the capacity to generate inhibitory signals via ά motor neurons to the antagonist muscle (Katz et al., 1991). This connection forms the reciprocal-inhibition reflex which facilitates the inhibition of the antagonist muscle (Enoka, 2002). The inhibition allows the activity between agonist and antagonist muscles to produce a more meaningful response in the muscle in which the stimulus (the stretch) was sensed (Enoka, 2002). Figure 2.5 is a representation of the reciprocal-inhibition reflex using the quadriceps and hamstrings muscles as agonist and antagonist respectively.

Figure 2.5: Neural circuits of the stretch reflex and reciprocal-inhibition reflexes (Enoka, 2002)

.



1 Ia afferents branch and synapse with an Ia inhibitory interneuron.

2 Stretch reflex innervating the agonist muscle (quadriceps femoris).

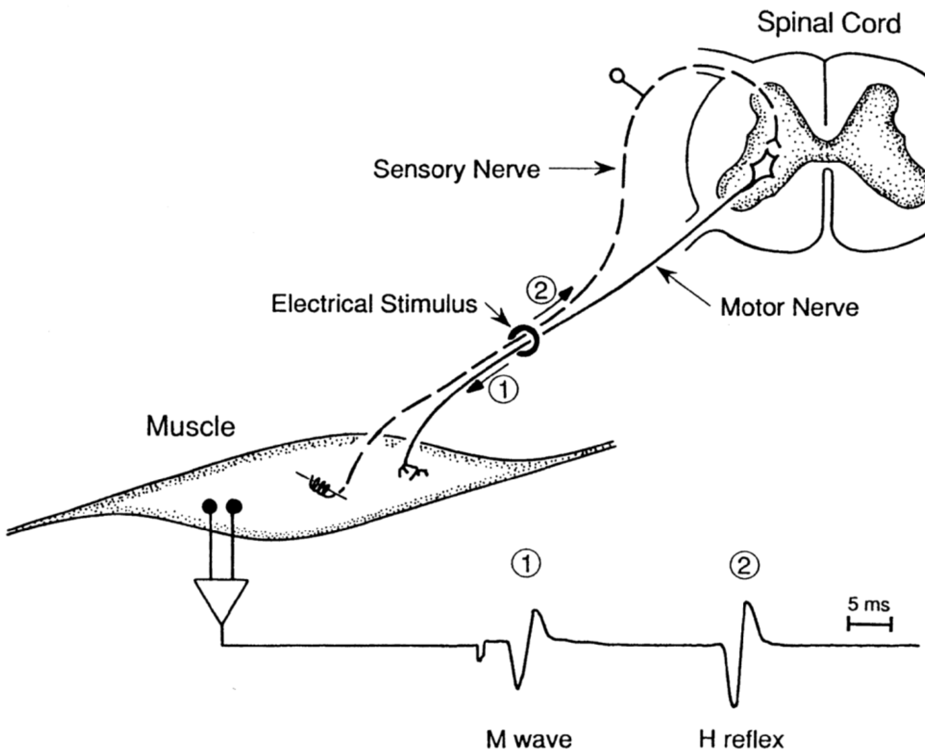
3 The antagonist muscle (hamstring) is inhibited allowing a more meaningful contraction of the quadriceps femoris.

2.3.1.4 Hoffmann reflex

The following introduction to the Hoffmann reflex (H-reflex) is summarised from Knikou (2008). The H-reflex is an electrically-induced monosynaptic stretch reflex which is evoked through low intensity electrical stimulation of an afferent nerve (Enoka, 2002), Figure 2.6. It has been suggested to be one of the most studied reflexes, which bypasses muscle spindles and therefore can be associated as a measure of Ia afferent sensitivity.

Figure 2.6: The Hoffmann reflex (Enoka, 2002)

.



1	M-wave, characterised by the direct motor response due to motor axons stimulation, detected before the H-reflex.
2	H-reflex, activation of group Ia afferent via electrical stimulation, propagated centrally to the spinal cord, where a postsynaptic potential is generated in the motor neuron, causing a contraction which is detected after the initial M-wave.

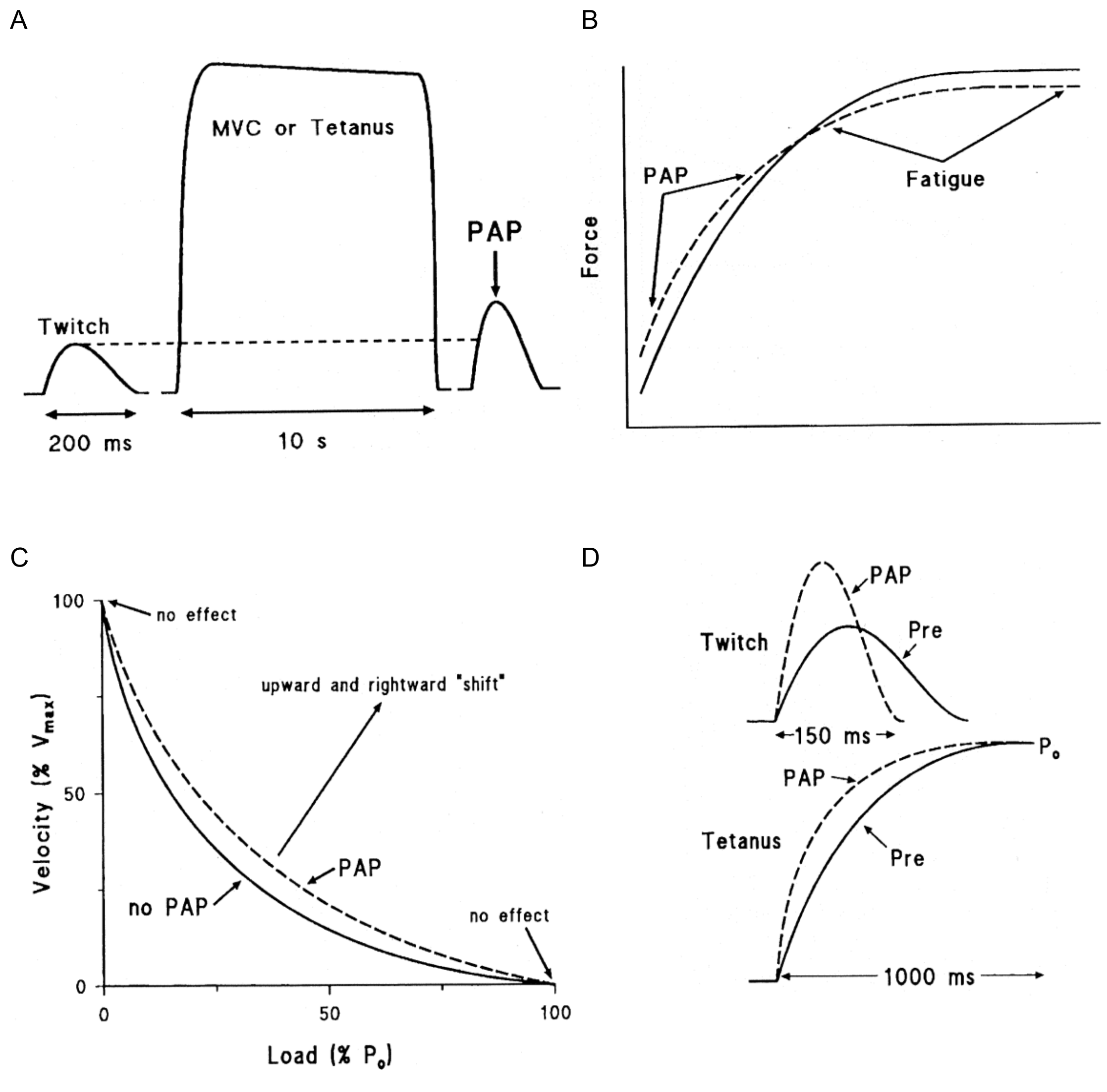
The motor response (M-wave) and H-reflex (see Figure 2.6) recruit different motor neurons, for example the M-wave involves activation of large diameter axons, innervating fast motor units. Whereas, the H-reflex involves small axons which innervate slow motor units. It is widely accepted that the H-reflex is a measure of motor neuron pool excitability resulting from a monosynaptic pathway (Figure 2.6). However, multiple synaptic inputs have sufficient time to contribute towards the H-reflex (Knikou, 2008).

It appears that the earliest portions of the H-reflex remain monosynaptic in nature, with later H-reflex components involving oligosynaptic pathways (relatively few synapses in series) (Misiaszek, 2003). These authors were careful to highlight that electrical stimulation within the H-reflex may not involve isolated recruitment of sensory nerve as perhaps simplistically indicated in Figure 2.6. The presence of a M-wave may indicate involvement of additional sensory afferents such as group Ib and II originating from Golgi tendon organs (GTO) and muscle spindles respectively (Misiaszek, 2003). The authors supported Knikou's (2008) view regarding the H-reflex as a measure of motor neuron excitability. However; Misiaszek (2003) highlighted that pre-motor neuronal events may also determine H-reflex characteristics. As a consequence H-reflex recordings should be carefully controlled during voluntary, sustained contractions and not at rest; due to significant intra- and inter-individual variability in excitability levels (Knikou, 2008).

2.3.1.5 Postactivation potentiation

An emerging candidate that may contribute towards a beneficial response following WBV is PAP (Cochrane et al., 2010). Therefore, the physiology and likely mechanism of the PAP phenomenon will be introduced in this section. PAP is categorised by an increase in muscle force following a "conditioning activity" as a result of contractile history (Sale, 2002; Robbins, 2005). The conditioning activity is typically electrically evoked contraction or a sustained maximal voluntary contraction (MVC), (see A, Figure 2.7). The resultant twitch following the conditioning activity is characterised by increased force capacity and shorter contraction time course.

Figure 2.7: Postactivation potentiation (PAP). A, an example of PAP following a 10s contractile activity. B, the effect of PAP on isometric force – frequency of stimulation, the induced PAP (dotted line) increases low frequency force, but not high frequency force. C, hypothesised effect of PAP on load (force) – velocity curve, PAP (dotted line) upward and rightward shift. D, the effect of PAP (dotted line) on force production during twitch and tetanus induced contraction, increasing rate of force development (Sale, 2002).



The majority of PAP literature has focussed on the influence on isometric contractions and is generally accepted that PAP raises the force output at lower motor unit firing frequencies (see B, Figure 2.7) (Sale, 2002). The influence of PAP on the load – velocity curve is minimal at both maximal velocity and load (see C, "no effect", Figure 2.7). However, PAP may increase the velocity of contractions, therefore inducing an upward and right shift (Sale, 2002). A consequence of increased velocity would likely be an increase in acceleration attained at these loads, which in turn would increase the rate of force development (RFD), (see D, Figure 2.7) (Sale, 2002). It appears the PAP may be greater in type II muscle fibres due to the fast type fibre characteristics which involve greater phosphorylation of myosin regulatory light chains (Sweeney et al., 1993). For a fuller account of PAP physiological characteristics, the reader is directed to the reviews of Hodgson *et al.* (2005); Robbins (2005); and Sale (2002).

The exact mechanisms to explain PAP remain unclear, the most commonly cited involves phosphorylation of myosin regulatory light chains (Sale, 2002; Sale, 2004; Hodgson *et al.*, 2005; Robbins, 2005). This is often known as twitch potentiation (Hodgson et al., 2005) which renders interactions of actin-myosin more sensitive to calcium (Ca^{2+}) release from the sarcoplasmic reticulum (Sale, 2002; Hodgson *et al.*, 2005; Cochrane *et al.*, 2010). As a result, there is a greater rate of cross bridge attachment for the same given concentration of Ca^{2+}, which increases twitch tension (Zhi *et al.*, 2005; Cochrane *et al.*, 2010). For an in-depth biochemical review see Sweeney *et al.* (1993). Another mechanism to account for PAP has been suggested as increases in α-motor neuron excitability via changes in H-reflex (Hodgson et al., 2005). This has already been outlined in section 2.3.1.4 and will also be dealt with in section 2.3.3.

2.3.2 The tonic vibration reflex

One of the first candidates put forward to explain the effects of vibration was the TVR. This phenomenon stemmed from work by Eklund & Hagbarth (1966). These authors reported that mechanical vibration applied to a skeletal muscle tendon evoked an involuntary tonic reflex muscle contraction. It was stated that the TVR may be elicited in all skeletal muscle (Eklund & Hagbarth, 1966). The strength of TVR response was found to be proportional to the frequency of vibration (Eklund &

Hagbarth, 1966). The authors argued the TVR is dependent upon the excitation of primary muscle spindle endings. Matthews (1966) proposed the reflex was monosynaptic and spinal in origin, based on animal studies in which a "classic stretch reflex" was elicited from decerebrated cats.

Burke *et al.* (1976a) supported the link between excitation of muscle spindle endings and the TVR. However, Burke *et al.* (1976a) found that a sharp response rate characteristic of muscle spindles did not match the gradual response rates of the TVR. Therefore, the TVR could not be explained through purely spinal reflex mechanisms, suggesting a central process or supraspinal mechanism (Burke et al., 1976a). More recent work has supported this and suggested the neuromuscular response to vibration may be mediated by monosynaptic and polysynaptic pathways (Cardinale & Bosco, 2003; Pollock *et al.*, 2012). This will be discussed in section 2.3.3.

Evidence supporting the theory of TVR during WBV exposure suggests that motor unit firing may be phase-locked to the WBV cycle (i.e. frequency); which could signify the motor unit activity is reflexive (Pollock et al., 2012). The authors proposed that if the motor unit activation response to WBV was postural or voluntary then firing patterns would be random, however all motor units were found to have the same reflexive type pattern.

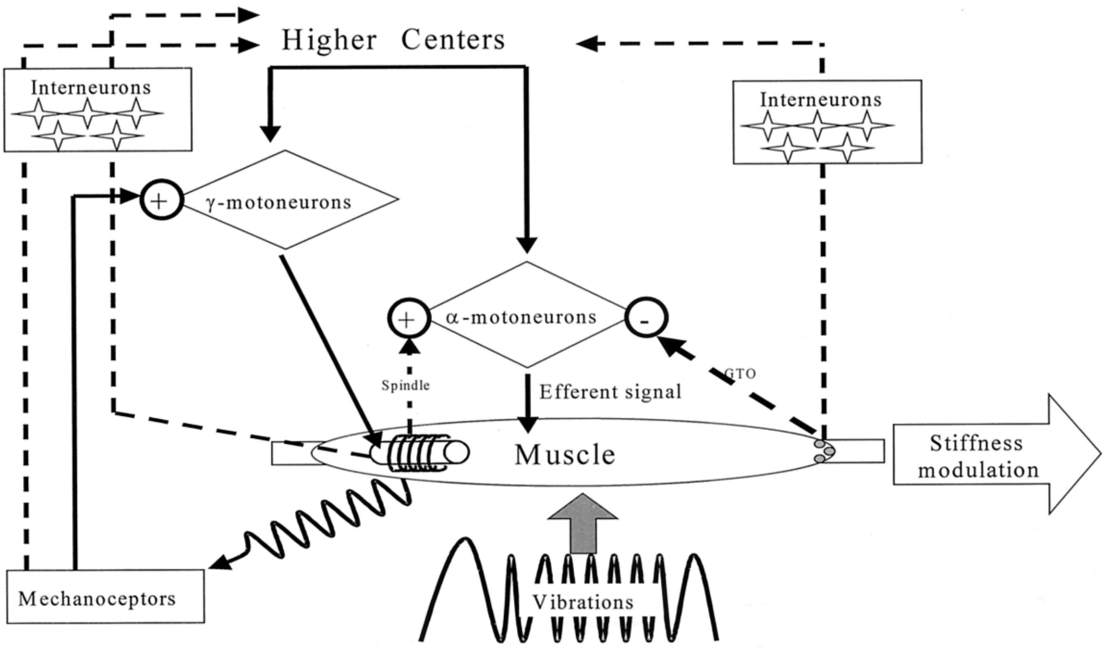
Nevertheless, there has been some debate regarding the TVR. Norlund & Thorstensson (2007) argued that the TVR and WBV are rarely discussed and lacked demonstration. Utilising strict inclusion criteria for their review paper, the link between the TVR and WBV appears tenuous (Nordlund & Thorstensson, 2007). It was argued that the work by Eklund & Hagbarth (1966) was based on a short exposure of high frequency vibration, directed specifically onto a tendon. The exposure of vibration as delivered by way of WBV involves longer exposure time and lower frequencies; especially at the target muscle groups where the magnitude of WBV may be lower, due to potential dampening (Nordlund & Thorstensson, 2007). Luo *et al.* (2008) suggested that following WBV, maximal knee extension may be associated with different mechanisms. This is supported by Pollock *et al.* (2010) as there has been reported limited response of thigh musculature EMG activity to frequency. This suggests that the TVR was not responsible for the increase in EMG activity associated with WBV (Pollock et al., 2010).

2.3.3 The role of mechanoreceptors

Over the last five to ten years further work developing the TVR mechanisms and the effect of vibration has involved muscle spindle activity (Adams et al., 2009). It has been suggested that the vibration stimulus may elicit small but rapid changes in muscle length (Cardinale & Bosco, 2003). This vibration induces muscle spindle activation (Kasai et al., 1992) and results in an increase of the muscle spindle and ά motor neurons inflow; effectively activating the stretch reflex via the Ia neuron loop, remaining a monosynaptic spinal reflex, Figure 2.4 (Bosco et al., 1999a). Cardinale & Bosco (2003) developed this proposed muscle spinal reflex mechanism, suggesting that vibration causes soft tissue deformation. During vibration excitatory inflow associated with the stretch reflex loop causes an increase in ά motor neuron activity through muscle spindle activation; which in turn causes a neuromuscular response via reflex muscle activity (Cardinale & Bosco, 2003). This would cause an enhancement of the stretch reflex loop via enhanced muscle spindle sensitivity and subsequent modulation of muscle stiffness, Figure 2.8 (Cardinale & Bosco, 2003).

Figure 2.8: The role of mechanoreceptors as a WBV mechanism to explain the modulation of muscle stiffness. GTO, golgi tendon organ; +, synaptic excitation; -, synaptic inhibition (Cardinale & Bosco, 2003).

.



2.3.3.1 Evidence supporting the role of mechanoreceptors

Cochrane *et al.* (2009) aimed to illustrate the muscle lengthening theory by directly measuring contractile tissue displacement via an ultrasound probe during WBV. Low WBV frequencies compared to a control (6 versus 0 Hz) resulted in significantly higher amplitudes of displacement for both musculotendinous complex length (375 versus 35 μm respectively) and contractile tissue (176 versus 4.2 μm respectively). However, the low frequency of WBV was required due to technical difficulties recording displacement via ultrasound at higher frequencies. Moran *et al.* (2007) suggested acute vibration may increase the signal size of the muscle spindle afferent. This may result in an increase in the sensitivity of the muscle spindle afferents to stretch (Moran et al., 2007). Costantino *et al.* (2006) speculated that the vibration induced stretch may cause an immediate reflex neurological response, stimulating the ά motor neurons located in the region of the muscle spindle; stemming from work related to the TVR and Burke *et al.* (1976a). As a result of vibration, a reflex muscle contraction aims to counteract the change in muscle length (Cochrane *et al.*, 2009). Further support of this mechanism was highlighted by Delecluse *et al.* (2003) who reported the enhancement of CMJ performance post WBV, was due to an enhancement in the stretch reflex, a known component within CMJ (Linthorne, 2001).

There is a growing body of evidence that vibration elicits the TVR (Mileva et al., 2009). Enhancement of spindle sensitivity, gains in ά motor neuron input and increases in the γ reflex loop have been proposed (Adamo et al., 2002). The vibration is likely to be detected by secondary sensory endings, thus involving the γ system (Cardinale & Lim, 2003b); suggesting the involvement of higher motor cortex polysynaptic pathways, Figure 2.8 (Cardinale & Bosco, 2003). During voluntary contractions cortically originating pathways are utilised and the effect of vibration on these cannot be excluded (Bosco et al., 1999a). It has been suggested that changing the Ia afferent input may change the excitability of corticospinal pathways (Carson et al., 2004). The corticospinal excitability can be measured through transcranial magnetic stimulation (TMS) (see Ross *et al.* (2007)) and WBV has increased tibialis anterior corticospinal excitability (Mileva et al., 2009).

The effects of WBV do not appear to be restricted to the peripheral spinal mechanisms; but also involve those supraspinal mechanisms including corticospinal and intracortical processes (Mileva et al., 2009). This supports previous work suggesting vibration may cause adaptive plastic changes within the central nervous system (Fattorini et al., 2006). These changes may include activation of muscle spindles and polysynaptic pathways eliciting the TVR (Torvinen *et al.*, 2002a; Ronnestad, 2004). The effect of WBV on the lowest and highest motor unit thresholds has been suggested to vary, which may explain these additional pathways of the TVR (Pollock et al., 2012). An increase in patella tendon reflex amplitude following WBV suggested that, WBV stimulus may cause stretch reflex changes, possibly due to central motor excitability (Rittweger et al., 2003). Conversely, prolonged vibration has been suggested to decrease cortical excitability (Issurin, 2005).

Muscle spindles may not be the only mechanoreceptors associated with vibration stimulus (Bosco et al., 1999a). Vibration below 40 Hz is thought to evoke a response from other mechanoreceptors such as Meissner's corpuscles (Hamano *et al.*, 1993; Zelena, 1994). Vibration is likely to be detected by the skin and other mechanoreceptors such as GTO, Figure 2.8 (Cardinale & Bosco, 2003). Work investigating the effect of vibration on short latency responses has suggested that if vibration were to increase spindle sensitivity, this could enhance the short latency responses (Melnyk *et al.*, 2008; Hopkins *et al.*, 2009). However, there has been an absence of effect detected, suggesting activation of other mechanoreceptors during vibration exposure (Melnyk *et al.*, 2008; Hopkins *et al.*, 2009).

2.3.3.2 Evidence against the role of mechanoreceptors

There is debate regarding the effect of vibration on muscle spindle sensitivity. Abercromby *et al.* (2007a) imposed a complex series of assumptions regarding muscle spindles and the Ia afferent input. This is summarised as follows: if the neuromuscular response to WBV were modulated by the Ia afferents; then the sensitivity of these would influence the magnitude of the muscle EMG activation during WBV. However, the results suggested increasing Ia afferent sensitivity had no effect on muscle EMG (Abercromby et al., 2007a). Limiting the study was a very short WBV exposure (15 s) and it remains unclear how many repetitions participants were exposed to. More importantly, based on the assumptions made by the authors, the findings should be viewed in the context of the study design which involved posture-specific squatting (40 º knee flexion).

Further debate exists, as there was inconclusive evidence whether the effect of vibration is due to changes in stretch reflex activity (Cochrane et al., 2008a). However, a number of limitations should be identified; the first was the use of the Jendrassik manoeuvre to comment on reflex activity. This isometric type movement involving clasping the hands together and attempting to pull them apart is thought to facilitate stretch reflex activity (Cochrane et al., 2008a). However, it has been suggested this is an inefficient method to facilitate the TVR, often producing weak responses (Eklund & Hagbarth, 1966). There were also no other measures of reflex activity employed by the study and also no specific explosive performance measures. Hopkins *et al.* (2009) provided a rationale that if WBV enhanced muscle spindle sensitivity then the short latency response could be affected. This would likely cause an increase in pre-activation and result in a shortening of the electromechanical delay (EMD) of a specific reflex, (in this case the patella tendon tap reflex). WBV appears not to affect spindle sensitivity as an enhancement in sensitivity would result in lowering firing thresholds and thus reduce the short latency response; neither of which has been apparent (Melnyk *et al.*, 2008; Hopkins *et al.*, 2009).

Although, measuring a certain stretch reflex through artificial initiation (tendon tap) may not conclusively measure the true response of muscle spindles to WBV (Kipp et al., 2012). The EMG measures utilised by Hopkins *et al.* (2009) and Hopkins *et al.* (2008) consisted of assessing both vastus lateralis and vastus medialis; both of which have proportions attaching to the upper and lateral borders of the patella. It could be argued that not all fibres may be involved in the reflex activity. The study also failed to assess any muscular performance. Conflicting with these findings are Melnyk *et al.* (2008) who reported WBV increases in hamstring short latency response following WBV. Clearly there is still some conflicting literature regarding the exact mechanisms governing the proposed enhancement in the stretch reflex loop due to WBV.

2.3.4 Neuromuscular facilitation

A number of authors have proposed a link between enhanced stretch reflex and neuromuscular facilitation (Bongiovanni et al., 1990). Neuromuscular facilitation stems from previous work researching neurological adaptations following resistance training. Before discussing neuromuscular facilitation in terms of a response to WBV it is worthwhile summarising the neurological adaptations mentioned. These adaptations are due to external stimuli (load) and chronic resistance training programmes have been shown to increase EMG activity (Hakkinen et al., 1985). Associated with an increase in EMG is higher force development capacity (Hakkinen et al., 1985). It has been argued that higher EMG responses have been correlated with increases in strength capacities (Sale, 1988).

The neurological adaptations which may be responsible for an higher EMG activity are increases in: synchronisation (Sale, 1988); recruitment (Hakkinen et al., 1985); and coordination of motor units (Rutherford & Jones, 1986). It is these adaptations which result in increases in the force generating capacity (Sale, 1988). Delecluse *et al.* (2003) suggested that WBV may elicit similar neurological adaptations as those elicited from resistance training. These adaptations tend to be from acute WBV applications and in a more chronic setting, adaptations via muscular hypertrophy must not be over-looked. However, there is some debate regarding a potential neuromuscular facilitation following WBV as some studies have found no increase in EMG following WBV (Hannah et al., 2012).

2.3.4.1 Synchronisation

The induced increase in EMG activity following acute WBV has been linked to motor unit synchronisation (Cardinale & Lim, 2003b). Further evidence supports that initial increases in strength following WBV are attributed to increased synchronisation (Adams *et al.*, 2009; Lamont *et al.*, 2010b). As the association between TVR and increases in synchronisation during voluntary muscle contractions has been made (Jordan et al., 2005); the increased synchronisation may account for the TVR contraction increase following vibration exposure (Jordan et al., 2005). In addition, WBV increased synchronisation of the muscle spindle afferent firing. However, specifics of an optimal frequency of WBV were not given (Jordan et al., 2005). Post WBV CMJ power and velocity may be dependent on synchronisation before dynamic explosive performance (Lamont et al., 2010a).

Mischi & Cardinale, (2009) suggested that increased synchronisation was the main mechanism able to increase the force generating capacity beyond 80 % maximal voluntary contractions. However, this was based on superimposed vibration to the upper body. Nevertheless, Torvinen *et al.* (2002b) supports the view that, following WBV exposure, increased synchronisation may enhance neuromuscular activation. However the authors were careful to point to that this is one of several candidates that could explain the effects of vibration. In addition, synchronisation is difficult to consistently demonstrate.

2.3.4.2 Recruitment

It has been proposed that vibration may promote and maintain a more synchronous pattern of motor unit recruitment (McBride *et al.*, 2004; Adams *et al.*, 2009). Increased recruitment may account for an enhanced muscle performance via a sustained increase in the ability to maximally activate muscles (Nordlund & Thorstensson, 2007). Vibration influencing mechanoreceptors and proprioceptors may have an effect on force generation; most likely from enhanced feedback from these receptors leading to increased motor unit recruitment (Hopkins et al., 2009).

An increase in spindle sensitivity may mean a greater number of activated spindles during WBV, leading to an increase in motor unit recruitment (Ronnestad, 2004; Wilcock *et al.*, 2009). WBV may maintain recruitment of high threshold muscle fibres thus influencing the force generating capacity of the muscle (McBride et al., 2004). Increased recruitment of previously inactive motor unit and high threshold motor unit recruitment have been proposed (Issurin, 2005; Lamont *et al.*, 2010a). Vibration may in fact reduce the recruitment threshold of fast twitch motor units (Di Giminiani *et al.*, 2009; Pollock *et al.*, 2012). Mischi & Cardinale (2009) attempted to explain the increase in EMG activity following vibration by also suggesting motor unit recruitment pattern may be altered. This altered pattern may be a result of neuromuscular strategies aimed at dampening the WBV stimulus (Mischi et al., 2010).

However, Fattorini *et al.* (2006) argued that altered recruitment could be excluded as a potential mechanism, reporting no increase in maximal voluntary contraction following vibration was observed. It must be stated that this was following research using locally applied vibration and not WBV, nor using vibration applied via an exercise device as per Mischi *et al*. (2010). Nevertheless, Colson *et al.* (2009) utilised WBV and supports this as no increase in muscle activation was reported. Though this was following an electrically-evoked muscle activation protocol and not during an explosive performance measures such jumping. In addition, as MVC also decreased following WBV a fatigue response may have been elicited.

Kipp *et al.* (2012) further contributed towards excluding recruitment as a mechanism following WBV, reporting H-reflex intensity significantly decreased. The frequency component of the H-reflex (relating to motor unit recruitment) remained unchanged, suggesting WBV may not alter recruitment patterns (Kipp et al., 2012). However, the authors were careful to avoid generalisation across multiple WBV platform types and protocols. It should also be noted that the work was based on a 5 minute continuous WBV protocol, and due to common methodological difficulties the H-reflex was obtained during "quiet standing". This may not reflect a true measure of H-reflex and motor neuron excitability (see section 2.3.1.4) (Knikou, 2008).

2.3.4.3 Coordination

The final component of neuromuscular facilitation has been proposed as an increase in coordination (Adams et al., 2009). During a chronic WBV programme early neural adaptations including a more complete motor unit activation, would involve an increase in coordination and lead to greater force generating capacity (Torvinen et al., 2002a). This increase in coordination aims to dampen the vibration stimulus (Torvinen et al., 2002a).

2.3.5 Hyper-gravitational forces

To understand the potential impact and mechanism that hyper gravitational forces may have, it is worthwhile discussing what effect hypo- or micro-gravitational forces have on muscle characteristics. It is widely accepted that immobilisation has the potential to be detrimental to neuromuscular performance (Fitts et al., 2001). Most commonly this type of research involves bed-rest studies in which muscular atrophy and deconditioning in both leg and spinal muscles have been observed (Blottner *et al.*, 2006; Belavy *et al.*, 2008). Micro gravity environments, such as spaceflight, have been shown to cause reductions of 13 % in calf muscle cross-sectional area (CSA) (Trappe et al., 2009) and accompanied by reduced force generating capacity (Lambertz et al., 2001). This was observed even when astronauts took part in regular resistance training. It appears vastus lateralis and soleus muscles are more susceptible to atrophy causing a reduction in maximal power capacity following time in microgravity environments (Fitts et al., 2001). This is likely due to the postural role these muscles have under normal gravitational conditions.

Gravity has been suggested to provide a large proportion of the mechanical stimulus required for the development of muscle structure (Bosco et al., 1984). Therefore, it would be logical to assume hyper-gravity environments could influence muscle mechanics. Based on the premise that, if a microgravity stimulus, through either bed rest or spaceflight causes atrophy; then hyper-gravity stimulus may positively influence muscle characteristics by increasing CSA and/or strength and power generating capacity. Bosco *et al.* (1984) found that an artificial hyper-gravity load (weighted body vests of 13 % body mass) enhanced explosive power performance versus a comparable normal gravity training programme. It is well established that vibration exposes participants to a hyper-gravitational stimulus due to the accelerations associated with WBV (Cardinale & Bosco, 2003; Wilcock *et al.*, 2009). Studies utilising accelerometers during WBV reported exposures ranging from 2 to 6 g (Cook et al., 2011).

It has been proposed that chronic VT induces similar adaptations to those that occur during chronic strength training programmes (e.g. muscular hypertrophy via exposure to external loads which induce structural muscular changes); and that a possible reason for this would be the exposure to an increased gravitational load (Cardinale & Bosco, 2003). Torvinen *et al.* (2002b) also highlighted that changes in morphological muscle structure over a 4 month WBV programme, may be due to the hyper-gravitational WBV stimulus. The increased load from hyper-gravitational forces could cause local structural adaptations resulting in hypertrophy (Nordlund & Thorstensson, 2007). Most commonly adopted postures during WBV consist of squats to a varying degree. During the posture the vibration superimposed by the platform would increase the load exposed to the muscle. This would increase muscle activation compared to non vibration squatting, leading to greater muscular adaptation (Wilcock et al., 2009). However, the same review highlighted that evidence is lacking for changes in muscle CSA and fast fibre muscle characteristics, as few studies have utilised magnetic resonance imaging (MRI) scans or muscle biopsies.

One such study did utilise MRI scans to assess spinal muscle atrophy but during a bed rest protocol study (Belavy et al., 2008). After 56 days bed rest a WBV group significantly increased psoas CSA versus a control bed rest group. Multifidus muscle atrophy was reported in both groups, although WBV attenuated this atrophy. However, it is important to note this study used spinal CSA and did not include any lower limb muscle groups. The direct link between spinal muscle and explosive performance for example may be limited. Further support for the positive effect of hyper-gravitational loads during WBV was provided by Blottner *et al.* (2006) who did investigate lower limb musculature. An increase in soleus CSA (% change vibration versus control of: +20 % versus -10 % for Type I fibres; and +40 % versus +12 % for Type II fibres) was observed following WBV during bed rest (56 days). There is a lack of research investigating the potential beneficial effects of hyper-gravitational forces during potential hypertrophy periods (i.e. in healthy active participants during training protocols).

2.3.6 Muscle tuning

An emerging candidate mechanism for the neuromuscular effect of vibration is muscle tuning. During physical activity the body is exposed to repetitive impact forces, such as running. As a result of heel strike, shock waves travel through the body, which are detected by many sensory receptors (Nigg & Wakeling, 2001). These shock waves are transmitted to both bone and the soft tissue package (a group of muscles recruited during the particular activity) (Nigg & Wakeling, 2001). The major frequency component during a running heel strike is 10 – 20 Hz and, as bone has a natural frequency of 200 – 900 Hz it is unlikely that running would produce a resonance effect (Nigg & Wakeling, 2001). As the natural frequencies of the soft tissue package is 5 – 65 Hz, the major frequency of impact matches closely to the natural frequency (Wakeling et al., 2002). As a consequence a resonance phenomenon may occur (Nigg & Wakeling, 2001) which is characterised by a greater D_{PTP} parameter of vibration (Wakeling et al., 2002). However, it has been established that muscles dampen this vibration quickly, thus avoiding a possibly problematic tissue resonance effect (Wakeling et al., 2002).

This muscular response to the excitation frequency of the impact force has been referred to as muscle tuning (Wakeling et al., 2002); which aims to minimise soft tissue vibration and is likely to be individual-specific and dependent on several characteristics of the soft tissue package (Nigg & Wakeling, 2001). The main three dependent variables appear to be: change in leg geometry; joint stiffness at ground contact; and changes in the wobbling mass model (Wakeling & Nigg, 2001).

Soft tissue, during heel strike and subsequent shock wave transmission, may act as a wobbling mass causing soft tissue to vibrate with a dampening pattern (Wakeling et al., 2002). This is in response to the vibration caused by a mechanical excitation (such as heel strike) (Wakeling et al., 2002). This mechanism was summarised by Cardinale & Wakeling (2005) suggesting after the initial impact (e.g. heel strike) the soft tissues vibrate. The tissues would continue to oscillate at their natural frequency, however a decay in D_{PTP} occurs due to a dampening effect (Wakeling *et al.*, 2002; Cardinale & Wakeling, 2005).

As the highest level of muscle activity has been found when vibration dampening occurs at the natural frequencies of the soft tissue; the dampening effect appears to be largest when the WBV frequency is closest to the natural resonant frequency of the soft tissue (Cardinale & Wakeling, 2005). It has been suggested that the wobbling mass theory would be most effective when these two frequencies were similar (Wakeling & Nigg, 2001). The dampening effect of the tissues may highlight a strategy to minimise any potential harmful effects during WBV (Cardinale & Wakeling, 2005).

Wakeling *et al.* (2002) suggested the dampening effect occurs due to the visoelastic properties of muscles. Mechanical energy is allowed to be stored through attached cross bridges within the muscle unit. If the muscle is in a rigid state the energy associated with the vibration is not absorbed; whereas once the muscle is in an activated state, the detachment and cross bridge cycling can absorb the energy from vibration (Wakeling et al., 2002). This dissipation of mechanical energy can be surmised as vibration dampening and may eliminate the oscillation (Cardinale & Wakeling, 2005).

WBV frequencies utilised within WBV literature are often within the natural frequency range and elicit the highest EMG response, e.g. (Cardinale & Lim, 2003b). Individual responses to WBV as mentioned by Di Giminiani *et al.* (2009) may be related to the individuals' vibration dampening ability, which may be due to the intrinsic muscle properties (Cardinale & Wakeling, 2005). Studies which have failed to show a beneficial effect of WBV on neuromuscular performance may not have targeted these muscles at their natural resonant frequencies. This may contribute to the conflicting results across the WBV literature.

2.3.7 Postactivation potentiation

The influence of the PAP phenomenon on performance is still emerging (see section 2.3.1.5). It is likely to have a beneficial influence on performance beyond the effects on isometric contractions as outlined in Figure 2.7 (Sale, 2002). The influence of PAP on ballistic performance still requires further research, nonetheless based on proposed PAP influences on RFD; jump performance may be enhanced (Sale, 2004). However, inconsistencies do exist across the literature (e.g. Bevan *et al.*, (2010) *cf.* Till & Cooke (2009)). This is likely due to variations in protocols used to elicit PAP in terms of repetition number, mode of exercise, recovery period etc.

The influence of PAP following WBV has only recently gained attention as a potential mechanism. Some of these have been referenced retrospectively in previous literature which demonstrated performance improvement post WBV (Bosco *et al.*, 2000; Torvinen *et al.*, 2002a; Cardinale & Lim, 2003a; Cochrane & Stannard, 2005). PAP has been anecdotally proposed as a mechanism to explain the response to WBV (Bazett-Jones et al., 2008a).

Cochrane *et al.* (2010) aimed to assess the acute effects of WBV on PAP via twitch potentiation measures. Five minutes of continuous Galileo WBV significantly increased both twitch peak force and RFD compared to a volume-matched control intervention up to 90 s post WBV. The authors argued that PAP was induced compared to a volume-matched posture without WBV stimulus. The type of WBV should be highlighted as Galileo is often restricted to below 30 Hz (26 Hz in this case), as should the mode of eliciting the twitch as this was via electrically evoked protocols. Dabbs *et al.* (2011) suggested that CMJ performance improvements up to 4 minutes post WBV may be partially explained by PAP. However, this was not directly measured through EMG or twitch response measures.

Colson *et al.* (2009) did record twitch magnitude following WBV and utilised a volume-matched control intervention, reporting an increase in single-peak twitch may be explained by PAP. However, neither double-twitch torque or activation levels changed post WBV, suggesting a full potentiation may not have occurred. Improvements in isometric MVC following WBV may have elicited PAP but the exact mechanisms are elusive as RFD, EMG or motor neuron excitability (H-reflex : M-wave ratio) remained unaltered post WBV (McBride et al., 2010). It should be noted that the PAP mechanism concerning phosphorylation of myosin regulatory light chains was not directly assessed.

The work by McBride and co-workers does introduce the conflict that exists regarding PAP as a mechanism to explain the WBV response. PAP of resting muscle twitch was not induced following NEMES WBV in athletes; and other neuromuscular performance markers (voluntary activation and peak isometric knee extension torque) were reduced following WBV (Jordon et al., 2010). The authors noted that the specific assessment protocol may have induced fatigue and therefore diminished any potential PAP effect.

Subsequent literature with specific aims to investigate PAP following WBV have utilised sprint performance as an indirect indicator of PAP (Guggenheimer *et al.*, 2009; Ronnestad & Ellefsen, 2011). Both studies utilised the same WBV type and investigated identical WBV frequencies. Ronnestad & Ellefsen (2011) suggested the improvement in sprint performance following 50 Hz, and not 30 Hz, signified PAP was induced following higher frequencies. However, Guggenheimer *et al.* (2009) reported no improvement in sprint performance following 50 Hz suggesting little efficacy for WBV as a means of inducing PAP associated performance gains. One important difference between these two studies was the postures and exercises completed during WBV exposure. Ronnestad & Ellefsen (2011) utilised a 30 s dynamic squatting protocol whereas Guggenheimer *et al.* (2009) involved a 5 s high knee running protocol. It would appear the shorter exposure time and alternating leg nature of the Guggenheimer *et al.* (2009) protocol was insufficient stimulus to elicit a PAP response. Nevertheless, both studies failed to directly measure neuromuscular responses which could characterise PAP such as EMG capable of explaining the improvement in sprint performance. The performance based outcome measures of both studies are also open to the role of expectancy effect, concerning the individual's belief regarding the efficacy of an intervention (McClung & Collins, 2007) (see section 5.1).

It appears problematic to investigate the precise PAP mechanisms responsible with regards WBV due to the generalised nature by which the WBV stimulus is exposed to both many muscular and neuromuscular systems (Jordon et al., 2010). Further research is warranted.

2.4 ACUTE WHOLE BODY VIBRATION

For the purpose of this literature review, acute WBV has been defined as research into the effect either immediately, or within 1 to 2 hours, post exposure. General populations include healthy, injury-free participants and those undertaking regular recreational exercise but not as part of a structured strength and conditioning programme. The electronic databases of PubMed (1958 to June 2012) and SPORTDiscus (1949 to June 2012) were searched using the following terms: 'whole body vibration' and 'vibration training'. Hand searches of other relevant journal titles also took place. Only full text English language articles were included in the following literature review. For clarity, the acute effects of WBV on EMG will be discussed separately. The acute effects of WBV on strength and power will be grouped by platform type as there is evidence of platform-specific responses to WBV (Abercromby et al., 2007a). As there have been suggestions that WBV may have a greater effect on athletes (Issurin & Tenenbaum, 1999; Luo *et al.*, 2005a) it is appropriate to review WBV literature on these participants separately. Athletic populations will be defined as those with at least 1 year of structured strength and conditioning programme training, competing at a professional or national level of their sport. These studies will be presented in part B of relevant tables

2.4.1 Effect on electromyography

There is conflicting evidence of the effect of acute WBV on EMG in general populations, for example: Roelants *et al.* (2006) and Abercromby *et al.* (2007a) *cf.*

Hopkins *et al.* (2008) and Hopkins *et al.* (2009), Table 2.1, part A. But also in athletic populations, for example Cormie *et al.* (2006) *cf.* Cardinale & Lim (2003b) Table 2.1, part B.

Table 2.1 Part A: Summary of the acute effects of whole body vibration on electromyography (EMG) outcome measures of general population participants. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement (D_{PTP}); s, seconds; min, minute; ↑, increase at significance level ($P < 0.05$); EMG, electromyography; EMG_{rms}, electromyography root mean square; EMD, electromechanical delay.

Author	Participants	WBV device	WBV characteristics	EMG outcome measures	Control group	Effects (all percentages are significant)
Abercromby *et al*. (2007a)	n = 16 Age: 32 No shoes worn	Galileo	30 Hz 4 mm 1. Dynamic squat 5 ° - 40 ° knee flexion 2. Static squat 20 ° knee flexion Approx. 1 x 30 s exposure	Vastus Lateralis Lateral Bicep Femoris Lateral Gastrocnemius Tibialis Anterior All during WBV exposure	Control group completed identical exercise without WBV exposure	Percentage ↑ above EMG_{rms} baseline: 1. Dynamic squat: Vastus Lateralis ↑ 26 % Bicep Femoris ↑ 30 % Gastrocnemius ↑ 106 % Tibialis Anterior ↑ 57 % 2. Static squat: Vastus Lateralis ↑ 103 % Bicep Femoris ↑ 10 % Gastrocnemius ↑ 151 % Tibialis Anterior ↑ 328 %
Hannah *et al. (*2012)	n = 14 Age: 22 No shoes	Powerplate	30 Hz 4 mm 140 ° knee flexion 5 x 60 s exposure	EMD for: Vastus Lateralis Vastus Medialis Rectus Femoris Bicep Femoris During electrically evoked contraction pre and post WBV	Control group completed identical exercise without WBV exposure	No significant interaction for EMD or EMG activity Recovery from max EMD significant different WBV vs. control group
Hopkins *et al*. (2009)	n = 22 Age: 23 Shoes worn	Galileo	26 Hz 8 mm 30 ° knee flexion 5 x 60 s exposure 60 s recovery	Patellar tendon tap reflex pre and post WBV: Vastus Lateralis Vastus Medialis Calculated EMD Knee extension force	Control group completed identical exercise without WBV exposure	No significant difference in EMG amplitude, EMD or force output during patella tendon tap reflex following WBV

Author	Participants	WBV device	WBV characteristics	EMG outcome measures	Control group	Effects (all percentages are significant)
Hopkins *et al*. (2008)	n = 22 Age: 22 Unknown footwear	Galileo	26 Hz 8 mm 30 ° knee flexion 5 x 60 s exposure 60 s recovery	Peroneus Longus EMD and reaction time following ankle inversion perturbation pre and post WBV	Control group completed identical exercise without WBV exposure	No significant difference in EMG activity, EMD or reaction time following WBV
Pollock *et al. (*2010)	n = 12 Age: 31 No shoes	Galileo	5,10,15,20,25 and 30 Hz 5 and 11 mm 15 ° knee flexion 1 x 7s exposure	Soleus Gastrocnemius Anterior Tibialis Rectus Femoris Bicep Femoris Gluteal Maximus All during MVC knee extension pre and post WBV	No control group	All muscles except Rectus Femoris and Gluteal Maximus ↑ EMG activity following higher frequency ↑ EMG activity following higher D_{PTP}
Roelants *et al*. (2006)	n = 15 Age: 21 Unknown footwear	Powerplate	35 Hz 5 mm 55 ° and 90 ° knee flexion 1 x 20 s exposure	Vastus Lateralis Vastus Medialis Rectus Femoris All during MVC knee extension pre and post WBV	Control group completed identical exercise without WBV exposure	Percentage ↑ in EMG_{rms} WBV vs. control: Vastus Lateralis ↑ 92.5 % and 51.7 % Vastus Medialis ↑ 102.5 % and 59.0 % Rectus Femoris ↑ 115.1 % and 49.1 %

Part B: Summary of the acute effects of whole body vibration on EMG in trained athletes. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; ↑ ↓, increase and decrease at significance level ($P < 0.05$); EMG, electromyography; CMJ, countermovement jump.

Author	**Participants**	**WBV device**	**WBV characteristics**	**Outcome measures**	**Control group**	**Effects**
Cormie *et al. (*2006)	n = 9 Age: 19-23 Resistance trained Unknown footwear	Powerplate	30 Hz 2.5 mm 1 x 30 s 100 ° knee flexion	EMG vastus lateralis, vastus medialis and biceps femoris during CMJ pre and post WBV	Control group completed identical exercise without WBV exposure	No change in EMG during CMJ following WBV versus control
Wakeling *et al*. (2002)	n = 20 Age: 26 Athletic Unknown training history No shoes worn	Hydraulic platform	1. Pulsed 13.1 Hz 5 mm 1 x 0.5 s 2. Continuous 10-65 Hz 5 mm 1 x 3 s	EMG rectus femoris, biceps femoris, tibialis anterior and gastrocnemius Soft tissue vibration via accelerometers All during WBV	No control group Order for continuous WBV was randomised	Mean EMG frequency ↑ during WBV EMG spectra peaked when input frequency was close to natural frequency for both pulsed and continuous trials
Cardinale & Lim (2003b)	n = 16 Age: 24 Professional volleyball players Unknown footwear	NEMES	30 Hz 10 mm 40 Hz 10 mm 50 Hz 10 mm 1 x 60 s 60 s recovery 100 ° knee flexion	EMG_{rms} vastus lateralis during WBV	Control group completed identical exercise without WBV exposure	All frequency ↑ EMG vs. control 30 Hz ↑ EMG by 34 % vs. 0 Hz 30 Hz ↑ EMG by 20 % vs. 50 Hz 40 Hz ↑ EMG by 10 % vs. 50 Hz

At this point it is important to differentiate between literature which recorded EMG activity during WBV and literature which recorded the acute effects of WBV in relation to EMG activity. Parts A and B of Table 2.1 detail whether EMG recordings were taken during WBV or pre and post WBV. Given this, it may explain the conflict mentioned in the previous paragraph such as Abercromby *et al.* (2007a) *cf.* Hopkins *et al.* (2008). The former study recorded EMG during WBV whereas the latter investigated the acute effects of pre and post WBV. Similarly, the conflict between Cormie *et al.* (2006) *cf.* Cardinale & Lim (2003b) in athletic populations, may be explained by the fact the former study recorded EMG activity during CMJ pre and post WBV, whereas the latter recorded EMG activity during WBV.

There are few studies which have investigated the EMG response during vibration in general populations. As such it is unclear if EMG activity increases due to WBV compared to identical postures without WBV stimulus (Roelants et al., 2006). Nevertheless, significantly higher EMG activity during MVC knee extension was found following acute WBV versus equivalent control isometric postures (Roelants et al., 2006). As this was based on muscle activity during an unloaded, single joint isometric test; it could be argued that it may not be comparable to weight bearing stance involving multi joints. Abercromby *et al.* (2007a) supported the work by Roelants *et al.* (2006) and found EMG significantly increased during WBV.

Higher EMG activity (5 – 50 % of MVC) was found following Galileo WBV with a main effect for higher D_{PTP} and significant interaction effects of frequency at higher D_{PTP} (Pollock et al., 2010), Table 2.1. This was found for all muscles except rectus femoris and gluteal maximus. Overall the authors suggested acute WBV involving high D_{PTP} elicited greater EMG activity, the effect of which increased with WBV frequency (e.g. at 30 Hz increases of 6, 7, 20 and 21 % for biceps femoris, rectus femoris, soleus and gastrocnemius respectively). However, this was based on research investigating Galileo WBV, and thus involved WBV frequencies limited to 5 to 30 Hz. The posture adopted of 15 º knee angle may have limited potential WBV responses (Ritzmann et al., 2012). Finally, vastus lateralis EMG activity was not recorded therefore WBV responses in this muscle group is unknown.

Cardinale & Bosco (2003) have suggested that WBV may increase neuromuscular performance and even excitability. It is interesting that Armstrong *et al.* (2008) investigated a direct measure of motor neuron excitability, the H-reflex (for definition see section 2.3.1.4), of the soleus muscle recorded post 1 minute of WBV. All participants found an immediate decrease in H-reflex during the first 1 minute post WBV, suggesting a negative influence on the neural excitability. Nevertheless, two common responses resulted in an increase in H-reflex higher than baseline levels. The wide range of participant activity levels may explain the highly variable response of the H-reflex. In addition, participants wore shoes but it is unknown if these were standardised shoes – the effect of which in dampening vibration is unknown. Given the fact this study only measured the response of a soleus H-reflex, caution should be advised if comparing this study to other muscles which would be involved when standing on WBV platforms.

Conflicting these findings, Hopkins *et al.* (2008) and Cormie *et al.* (2006) found no difference in EMG activity following WBV, Table 2.1. There was no effect of WBV on EMG activity during CMJ or other neuromuscular excitability markers, e.g. electromechanical delay, (EMD) following ankle inversion perturbation. However, a number of factors may have influenced the findings such as the choice of peroneus longus muscle, which may not have been exposed to an adequate magnitude of stimulus to elicit neuromuscular changes (Hopkins et al., 2008). As well as the short WBV exposure time utilised by Cormie *et al.* (2006). In addition, the posture adopted on the WBV platform may not have targeted the peroneus longus muscle. The EMD methodology involved artificially induced ankle inversion whilst walking along a trapdoor walkway. This may have produced an anticipatory effect or a change in the natural gait pattern. Finally, the use of shoes may have also affected the vibration exposed to the participants in this study via a dampening mechanism.

Hopkins *et al.* (2009) and Hannah *et al.* (2012) utilised quadriceps EMG activity and EMD measures and again found no acute WBV effect, suggesting that the acute effect of WBV on EMG during both: electrically evoked and tendon tap reflex contractions respectively, is minimal. However, a number of issues should be highlighted such as, the use of an artificially induced stretch reflex which may not accurately measure overall neuromuscular excitability. Other measures (e.g. H-reflex) appear to recruit higher threshold motor units than tendon taps (Kipp et al., 2012). Participant inclusion criteria required a measurable patella tendon tap reflex. Therefore, those included in the study were at a certain level of reflex sensitivity prior to WBV exposure. It could be speculated that these participants may have reached a ceiling effect and using those who did not have a measurable reflex may have produced different results. Also that the nature of controlled and artificially-induced single jointed stretch reflexes may not reflect the more dynamic nature of stretch reflexes induced during performance.

Wakeling *et al.* (2002) supports the only other study (Cardinale & Lim, 2003b) to report increases in leg EMG activity during WBV, although Wakeling and co-workers used a hydraulic WBV device. Both pulsed and continuous WBV protocols reported increased mean EMG frequency. The increased EMG activity peaked when the WBV frequency matched the natural frequency of the soft tissue, (i.e. muscle tuning, see section 2.3.6). One potential limitation would be the posture adopted when standing on the WBV platform. This consisted of forefoot standing with heels off the platform and may have affected the gastrocnemius EMG data. This is likely due to reduced vibration transmission through the heels and therefore reduced exposure of the calf musculature to the vibration stimuli.

Cardinale & Lim (2003b) found that in female professional volleyball players, one 60 s repetition of 30 Hz NEMES WBV significantly increased EMG_{rms} activity by 34 % versus 0 Hz in the same posture. All WBV frequencies (30, 40, 50 Hz) significantly increased EMG activity versus 0 Hz. It seems that for female volleyball players 60 seconds of 30 Hz WBV elicits a beneficial effect for EMG measures at least. Volleyball players may have a certain level of neuromuscular adaptations associated with a stretch reflex during jumping activities. This may predispose volleyball players to gains post WBV, although there was a lack of direct strength or power performance measures.

Roelants *et al.* (2006) highlighted the question that if EMG activity was significantly higher would this equate to an increase in strength and power performance? It could be argued that increased EMG activity may signify increases in motor unit synchronisation, coordination and/or recruitment, mechanisms which have already been discussed (see section 2.3.4). It has been proposed that increases in motor unit characteristics may result in increases of strength and power generating capacity (Adams et al., 2009).

As shown in Table 2.1 the EMG response was highly variable and individual specific within four common responses. Di Giminiani *et al.* (2009; 2010) presented evidence of an individual EMG response to acute WBV frequency although, the individual EMG response was used within a chronic (8 weeks) WBV programme.

In summary, EMG responses to acute WBV in both general populations and trained individuals appear to be generally higher than without WBV, however conflicting evidence does exist. For example, there does not seem a WBV effect on neuromuscular excitability (e.g. EMD).

2.4.2 Effect of acute whole body vibration on strength and power

As already discussed (see section 2.2.2) there appears to be evidence of a platform-specific response. This is likely to be due to different WBV characteristics delivered by different WBV platforms, such as vibration direction (Abercromby *et al.*, 2007b, a). Therefore, it seems appropriate to group the acute responses to WBV by platform type. In trained athletes, the three common types of WBV: Powerplate; Galileo; and NEMES remain. On reviewing the literature there are additional types of WBV devices; iTONIC, Pneu-Vibe and Med-Vibe. At this point it is worthwhile acknowledging a body of literature using athletic participants and: locally applied vibration; vibration delivered by vibrating dumbbell; and superimposed vibration during exercise. As valuable as this body of research is, it is out of the remit of this literature review and therefore will not be discussed.

2.4.2.1 Powerplate whole body vibration

Those studies which investigated the effect of acute WBV from Powerplate platforms on strength and power are summarised in Table 2.2. Only three studies have utilised Powerplate in the general populations (Bazett-Jones *et al.*, 2008a; Adams *et al.*, 2009; Marin *et al.*, 2010), (part A) and only two studies utilised athletic participants (part B).

Table 2.2 Part A: Summary of the acute effects of whole body vibration delivered by Powerplate devices on strength and power parameters of general population participants. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; ↑ ↓, increase and decrease at significance level ($P < 0.05$); CMJ, countermovement jump; NS, non-significant.

Author	**Participants**	**WBV device**	**WBV characteristics**	**Outcome measures**	**Control group**	**Effects**
Adams *et al*. (2009)	n = 22 Age: 30 Unknown footwear	Powerplate	30-50 Hz 2-4 mm + 4-6 mm 45 ° knee flex 1 x 30 s, 45 s, 60 s	Leg power calculated from CMJ via contact mat	None Randomised Unclear if blinded	CMJ power ↑ 0.5 and 1.1 % above baseline scores at 1min and 5min post WBV Greatest power ↑ was at low frequency with low peak to peak displacements and high frequency with high peak to peak displacements
Bazett-Jones *et al*. (2008)	n = 44 Age: 20 Unknown footwear	Powerplate	30 Hz 2-4 mm 40 Hz 2-4 mm 35 Hz 4-6 mm 50 Hz 4-6 mm Squatting to 90 ° knee flexion 1 x 45 s	CMJ height via linear position transducer	Control intervention of 0 Hz and 0 mm Randomised Blinded	No significant difference in CMJ height in male participants 40 Hz 2-4 mm ↑ CMJ height by 9 % in female participants 50 Hz 4-6 mm ↑ CMJ height by 8 % in female participants
Marin *et al*. (2010)	n = 20 Age: 19 Shoes worn	Powerplate	30 Hz 1.15 mm 50 Hz 2.41 mm 30 ° knee flexion WBV during elbow extension	Elbow extension repetition number and velocity	Control intervention of 0 Hz and 0 mm	Repetition number and velocity significantly ↑ both 30 Hz and 50 Hz versus control

Part B: Summary of the acute effects of whole body vibration delivered by Powerplate, on strength and power parameters of trained athletes. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; ↑ ↓, increase and decrease at significance level (P < 0.05); CMJ, countermovement jump; rec, recovery; IMVC, isometric maximal voluntary contraction; NS, non-significant.

Author	**Participants**	**WBV device**	**Vibration characteristics**	**Outcome measures**	**Control group**	**Effects**
Cormie et al. (2006)	n = 9 Age: 19-23 Resistance trained Unknown footwear	Powerplate	30 Hz 2.5 mm 1 x 30 s 100 ° knee flexion	CMJ height and peak power via force plate IMVC squat peak force	Control trial completed identical exercise without WBV exposure	CMJ height as expressed as a % of baseline values ↑ post WBV by approximately 1 % NS difference in CMJ peak force and peak power NS difference in IMVC squat peak force
Lovell *et al.* (2012)	n = 10 Age: 20 Semi pro soccer players	Powerplate	40 Hz 0.8 mm 3 x 60 s 60 s rec Approximately 30 ° knee flexion At half time of 90 min simulated game	Sprint performance CMJ height	Control intervention involving seating Agility intervention	NS difference in sprint performance Significantly lower decrease in CMJ height following WBV versus control

Small but significant increases in CMJ performance has been found in: general populations of 0.5 to 1.1 % (Adams et al., 2009); and in athletic populations of 0.7 % (Cormie et al., 2006). In general populations these small increases would likely be negligible; however, increases of 0.7 % in the context of an athletic population may be meaningful. The authors did state the participants were athletes however, it was not stated the type, level or training history of the participants' sport. The other limitation with the work by Adams *et al.* (2009) is the lack of a volume-matched control group as the half squat posture adopted during the WBV stimulus may account for the small increases in CMJ power.

One such study which did utilise a control group was Bazett-Jones *et al.* (2008a). The control group was volume-matched squatting to 90 º knee flexion at identical squatting rate without WBV stimulus. To add strength to the methodology the study design was frequency-blinded and randomised. No effect for CMJ height was found in male participants. Whereas, 45 s of WBV (at both 40 and 50 Hz) increased female CMJ height by 9 and 8 % respectively. Even with the methodological strengths the work by Bazett-Jones *et al.* (2008a) does have limitations in methodological design, as discussed in section 2.2.3.1. In addition, the authors reported a range for D_{PTP} (2-4 and 4-6 mm), therefore the resultant accelerations would be expected as a range. Bazett-Jones *et al.* (2008a) utilised platform accelerations to quantify WBV acceleration, thus expressing acceleration as a single figure.

However, using the equation by Wilcock *et al.* (2009) a reported 21.2 $m.s^{-2}$ would correspond to an acceleration range of 25.1 – 50.2 $m.s^{-2}$. Clearly there is some discrepancy in the reported acceleration by Bazett-Jones *et al.* (2008a) and the theoretical accelerations. The type of footwear worn was not stated which could introduce a variable in WBV stimulus. A final limitation is the use of jump displacement data collected via a stick held on the participants' shoulders which may affect the accuracy of the data.

WBV during elbow extension significantly increased velocity and repetition number till failure (Marin et al., 2010). Limitations included: isometric lower leg WBV exposure and dynamic upper body strength outcome measures; authors admitting footwear worn dampen the WBV stimulus but not quantifying this; and finally the potential placebo issues due to the outcome measures having an element of voluntary motivation. This could be characterised in a role of expectancy effect. Lovell *et al.* (2012) reported an attenuation in CMJ performance decrease at half time of a simulated football game.

Overall there is a small amount of literature to suggest that strength and power increases following acute Powerplate WBV. This is more evident in jump performances of both general populations and trained, but the latter is very limited in the amount of available evidence. The literature has methodological issues regarding: lack of volume-matched comparisons; simultaneous changes in WBV parameters; footwear and potential dampening of the WBV stimuli; and potential role of expectancy effects (see section 5.2.3.6).

2.4.2.2 Galileo whole body vibration

Those studies which utilised a Galileo WBV platform are summarised in Table 2.3.

Table 2.3 Part A: Summary of the acute effects of whole body vibration delivered by Galileo devices on strength and power parameters of general population participants. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; ↑ ↓, increase and decrease at significance level ($P < 0.05$); MVC, maximal voluntary contraction; MRFR, maximum rate of force rise; CMJ, countermovement jump; D_{PTP}, peak to peak displacement; DF, dorsiflexion; PF, plantarflexion; BM, body mass.

Author	**Participants**	**WBV device**	**WBV characteristics**	**Outcome measures**	**Control group**	**Effects**
Cochrane *et al*. (2008b)	n = 8 Age: 28 No shoes worn	Galileo	26 Hz 6 mm Squatting to approx. 90 ° knee flexion at a tempo of 6 s	CMJ height via force plate	Control groups included cycle and warm bath All interventions completed to reach similar muscle temperature	WBV significantly ↑ CMJ height
Dabbs *et al. (*2011)	n = 30 Age: 24 Unknown footwear	Med-vibe (rotational)	30 Hz 6.5 mm ¼ squat 4 x 30 s 30 s recovery With varying rest intervals to post WBV tests	CMJ height	Volume-match control group	WBV significantly increased CMJ height regardless of rest intervals
De Ruiter *et al*. (2003a)	n = 12 Age: 23 No shoes worn	Galileo	30 Hz 8 mm 100 ° knee flexion 5 x 60 s 120 s recovery	MVC Knee extension MRFR	No control group Pre testing over 2 days to ↓ learning effect	90 s post WBV MVC ↓ significantly greater than MRFR Significant ↓ in MVC lasted until 60 min post WBV
Jacobs & Burns (2009)	n = 20 Age: 29 Rubber soled shoes	Galileo	0-26 Hz during 1st min then 26 Hz for 5 min Unknown D_{PTP} and posture	Knee extension and flexion torque via Biodex	Randomised order of WBV or cycling of a same duration	Average extension torque ↑ 7 % WBV vs. ↓ 0.7 % cycle Peak extension torque ↑ 9.6 % WBV vs. ↓ 2.6 % cycle

Author	Participants	WBV device	WBV characteristics	Outcome measures	Control group	Effects
Kemertzis *et al*. (2008)	n = 21 Age: 21 No shoes worn	Galileo	26 Hz 0 to 4.5 mm due to foot position 5 x 60 s 60 s rec 0 ° knee flexion 80 % BM load	Peak ankle DF/PF torque Angle of peak torque	Randomised cross-over design No control group of no WBV with 80 % BM load	No significant difference PF peak torque WBV caused a significant shift in angle of peak PF torque towards a longer muscle length.
Rittweger *et al*. (2000)	n = 37 Age: 23 Unknown footwear	Galileo	26 Hz 10.5 mm Squatting with 40 % BM till exhaustion	MVC force knee extension	Control intervention Bike till exhaustion	Knee extension force significantly ↓ WBV vs. bike
Rittweger *et al*. (2003)	n = 19 Age: 23 Unknown footwear	Galileo	26 Hz 6 mm Squatting down to 90 ° knee flexion till exhaustion with 40 % BM load	Serial jumping via contact mat	Randomised cross over design Control group completed identical exercise without WBV exposure	No significant difference in serial jumping

Part B: Summary of the acute effects of whole body vibration delivered by Galileo on strength and power parameters of trained athletes. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; m, metre; ↑ ↓, increase and decrease at significance level (P < 0.05); exs, exercises performed on the WBV platform; CMJ, countermovement jump; NS, non-significant; IMVC, isometric maximal voluntary contraction.

Author	Participants	WBV device	Vibration characteristics	Outcome measures	Control group	Effects
Bosco et al. (1999b)	n = 6 Age: 20 National level volleyball players Gym shoes	Galileo	26 Hz 10 mm 10 x 60 s 60 s recovery 100 ° knee flexion One leg exposed to WBV	Unilateral leg press 70, 90, 110, 130 kg Mean velocity, acceleration, force, power calculated	One leg acting as control Unknown if control leg completed the same posture with no WBV exposure	WBV ↑ mean power of all loads by 5.8 – 8.2 % respectively WBV ↑ mean velocity all loads by 5.2 – 7.9 % respectively Force velocity curve shifted to the right. NS difference in control leg
Cochrane & Stannard (2005)	n = 18 Age: 22 “Elite” hockey players Shoes worn	Galileo	26 Hz 6 mm 5 x 60 s of: 4 exs 2 x 30 s of: 2 exs	CMJ height via contact mat	Control group completed identical exercise without WBV exposure	WBV ↑ CMJ height by 8.1 % NS difference in control group
Crow *et al.* (2012)	n = 22 Age: 22 “Elite Aussie Rules” players Shoes worn	Galileo	30 Hz 6.4 mm 1 x 45 s 10 – 30 ° knee flexion	Loaded 20 kg CMJ power via linear encoder	Group performing gluteal muscle exs Control group no exercise	↑ CMJ following gluteal muscle exs versus WBV and control groups ↓ CMJ following WBV versus control group
Stewart et al. (2009)	n = 12 Age: 24 Trained participants Shoes worn	Galileo	26 Hz 4 mm Either 2, 4, 6 min continuous 24 hr. recovery 5 ° knee flexion	3 x 2s IMVC knee extension	None	2 min WBV ↑ peak torque by 4 % 4 min WBV ↓ peak torque by 3 % 6 min WBV ↓ peak torque by 6 %

Abercromby *et al.* (2007a) has already been discussed (see section 2.4.1) and will not be re-examined again in this section.

The acute effect of Galileo WBV on MVC knee extension appears, at first, to be detrimental. Rittweger *et al.* (2000) found MVC knee extension torque and SQJ height significantly lower post WBV. However, this was based on squatting during WBV exposures. More importantly, this squatting involved a 35-40 % BM load and an exhaustive protocol. The additional load whilst on the WBV platform may have influenced the actual vibration magnitude exposed to the participants. It is unknown how additional loads may change the vibration dose (Wilcock et al., 2009) (see chapter 7). In addition, the exhaustive protocol may explain the reduced knee extension torque. A further limitation of Rittweger *et al.* (2000) was a lack of a volume-match control group, as the control group performed exhaustive cycling. It is unknown if the significant decreases in strength and power performances were due to WBV or the exhaustive squatting.

Rittweger *et al.* (2003) addressed some of the aforementioned methodological weakness by using a randomised cross-over design and volume-matched control group. The authors did however continue using loaded squats during the WBV exposure. No significant difference was found in serial jumping performance. Again the exhaustive nature of the squatting protocol, along with the use of additional loads during WBV exposure, may have minimised any potential beneficial effect on strength and power performances.

Supporting the detrimental effect on strength and power is de Ruiter *et al.* (2003) who found significant declines in knee extension force post Galileo WBV. Again, there was a lack of volume-matched controls; therefore, it is unknown whether the WBV stimulus, or the posture maintained, caused the acute decrease in knee extension force. Conflicting with de Ruiter and co-workers are other studies which did utilise a volume-matched control group (Torvinen *et al.*, 2002a; Dabbs *et al.*, 2011). Significant increases in isometric strength and CMJ performance regardless of rest interval were found respectively. The findings by Dabbs *et al.* (2011) are limited by the use of a vertical jump tester measuring jump height via arm swing movements which was recorded to the nearest 7 mm.

The study also utilised an exercise protocol whilst standing on the WBV platform which involved squatting, jumping and alternating body weight leg to leg. These appear to have lacked standardisation and little joint angle and temporal information was given, making replication of the WBV protocol difficult. Furthermore, one of the activities involved jumping, clearly while in the air the participants would not be exposed to vibration stimulus. Equally as important, when the participants land it is unknown what effect the likely additional downward g-forces may have. Again, replication of this protocol is difficult as no information regarding recovery periods were given. Supporting Torvinen *et al.* (2002a) was Cochrane *et al.* (2008b) who reported acute increases in CMJ height post Galileo WBV, but they failed to utilise volume-matched controls.

Jacobs & Burns (2009) reported an increase in average and peak knee extension torque of 7.7 and 9.6 % post Galileo WBV respectively versus post cycling. This study is limited by the lack of a volume-matched control group and that no information was given on the D_{PTP} or posture adopted during WBV exposure.

A final study investigated the acute effect of Galileo WBV on strength and power parameters (Kemertzis et al., 2008). No significant differences in either dorsiflexion (DF) or plantarflexion (PF) peak torque were found. Although, there was a shift in PF angle of peak torque towards a longer muscle length. However, participants were instructed to maintain only 80 % of their body mass (BM) on the WBV platform. Participants were instructed during familiarisation, using electronic scales to monitor BM on the WBV platform; but during WBV exposure these scales were removed. As a result participants may have placed less body weight on the platform, theoretically influencing the magnitude of WBV exposure. The posture adopted of 0 º knee flexion is likely to have affected the potential response to WBV as explained earlier (Cardinale & Lim, 2003b). Overall the effect of acute Galileo WBV on general populations seems to be conflicting with several study limitations identified.

Bosco *et al.* (1999b) found both average power and average velocity increased for a range of leg press loads post Galileo WBV in national level volleyball players. As a methodological strength one leg acted as an experimental leg, the other acted as a control leg. However, it is unknown if the control leg adopted the same 100 º knee flexion on the WBV platform without receiving any WBV stimulus. What is also unknown is the effect on one leg standing whilst receiving the WBV stimulus. It is conceivable that the experimental leg would have a higher muscle activity due to the one leg stance and this may have resulted in the increases in strength and power performances. Nevertheless, in these highly trained explosive athletes WBV appears to have caused a shift to the right in the force velocity curve. Whether this would necessarily equate to an increase in performance is debatable. Although, isometric voluntary contraction (IMVC) of knee extension increased following 2 minutes Galileo WBV of the same frequency (Stewart et al., 2009). Albeit this study lacked volume-matched controls

Cochrane & Stannard (2005) utilised a volume-matched control and the same frequency of WBV (26 Hz), showing improved CMJ height by 8.1 % in elite hockey players. However, there are a number of limitations within this study. The participants wore shoes which may have had a dampening effect; as would performing an exercise protocol during WBV, for the reasons mentioned previously. The CMJ outcome measure included arm initiated movement which is problematic to quantify the potential affect that may have had on CMJ performance.

Conflicting with the work by Cochrane & Stannard were Crow *et al.* (2012) who reported a detrimental effect on CMJ power following WBV versus both control and gluteal exercises. However, WBV exposure time was only 45 s and compared to seven sets of gluteal exercise each of 10 repetitions, it is unlikely these are volume-matched. The knee flexion angle of 10 – 30 º is unlikely to place the quadriceps muscle group under stretch. This a muscle group likely to be influential in CMJ performance and perhaps not targeted for WBV stimulus, reducing any potential response (Ritzmann et al., 2012).

As with Powerplate, Galileo responses to acute WBV appear to be beneficial to jump performance. For other strength and power measures in general populations literature suggests that there is no change or in fact may actually decrease performance. Again, similar to section 2.4.2.1, a lack of volume-matched controls exist but also the use of exercise protocols during WBV may have an unknown effect on the received WBV stimuli.

2.4.2.3 NEMES whole body vibration

Those studies which utilised NEMES WBV are summarised in Table 2.4. Also included is a study (Torvinen et al., 2002c) which utilised a similar vertical oscillating WBV platform (Kuntotary), in general population participants, (part A).

Table 2.4 Part A: Summary of the acute effects of whole body vibration delivered by NEMES and Kuntotary devices on strength and power parameters of general population participants. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; ↑ ↓, increase and decrease at significance level ($P < 0.05$); SQJ, squat jump; CMJ, countermovement jump; 1RM, one repetition maximum; IMVC, isometric maximal voluntary contraction.

Author	Participants	WBV device	WBV characteristics	Outcome measures	Control group	Effects
Cardinale & Lim (2003a)	n = 13 Age: 21 Unknown footwear	NEMES	High Hz: 40 Hz 4 mm Low Hz: 20 Hz 4 mm Semi squat 5 x 60 s 60 s recovery	SQJ CMJ	None	Low Hz: ↑ 3.9 % SQJ height High Hz: No significant difference in SQJ height High Hz: No significant difference in CMJ height Low Hz: No significant difference in CMJ height
Da Silva-Grigoletto et al. (2006)	n = 31 Age: 20 Unknown footwear	NEMES	20,30,40 Hz 4 mm Unknown posture 6 x 60 s 120 s recovery	SQJ height via infra red platform CMJ height via infra red platform Average velocity, average power & 1RM during concentric phase of 90 º knee flexion squat	Order randomised No control group Unknown if subjects were blinded to frequency	20 and 30 Hz: ↑ 3.7 and 3.5 % SQJ height respectively 40 Hz: ↓ 2.5 % SQJ height 20 Hz: No significant difference in CMJ height 30 Hz: ↑ 4.6 % CMJ height 40 Hz: ↓ 2.7 % CMJ height 1RM: No significant difference for any frequency 20 and 30 Hz: ↑ 1.5 and 4.6 % average squat power respectively 40 Hz: No significant difference for any frequency
Torvinen et al. (2002c)	n = 16 Age: 18 – 35 Thin gym shoes	Kuntotary	4 minutes WBV with 4 x 60s components at 25 Hz, 30 Hz, 35 Hz and 40 Hz 2 mm Exs programme during WBV exposure	IMVC knee ext CMJ height via contact platform Shuttle run	Crossover design acting as their own control group completing identical exercise without WBV exposure	IMVC, CMJ height, or shuttle run no significant differences

Part B: Summary of the acute effects of whole body vibration delivered by NEMES on strength and power parameters of trained athletes. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; m, metre; ↑ ↓, increase and decrease at significance level ($P < 0.05$); SQJ, squat jump; CMJ, countermovement jump; NS, non-significant; MVC, maximal voluntary contraction.

Author	Participants	WBV device	Vibration characteristics	Outcome measures	Control group	Effects
Bullock *et al.* (2008)	n = 7 Age: 25 International skeleton athletes Unknown shoes and socks	NEMES	30 Hz 4 mm 3 x 60 s with 1:3 ratio for work : recovery 110 ° knee flexion	SQJ and CMJ via linear encoder around waist belt 1 x 30 m sprint time using laser timing device	Control group stood on WBV plate in upright stance with no vibration	NS difference in SQJ performance NS difference in CMJ performance NS difference in 30 m sprint time Trend towards ↑ velocity during sprints post WBV (P = 0.25)
Bullock *et al.* (2009)	n = 4 Age: 22 International skeleton athletes Shoes worn	NEMES	45 Hz 4 mm 3 x 60 s 110 ° knee flexion	1 x 30 m sprint time using laser timing device Skeleton push sprint time	Volume-matched control	NS difference in 30 m sprint time NS difference in skeleton push sprint time
Jordon *et al.* (2010)	n = 24 Age: 28 National or college athletes	NEMES	30 Hz 4 mm 3 x 60 s 130 ° knee flexion	Peak torque during isometric MVC knee extension	Volume-matched control Counter-balanced study design	Peak torque ↓ following both WBV and control Peak torque ↓ was less following WBV versus control (↓ 1.9 and ↓ 8.9 % respectively)

To date there are only two studies investigating the effect of acute NEMES WBV on strength and power parameters in general populations (Cardinale & Lim, 2003a; Da Silva-Grigoletto *et al.*, 2006). The first study reported SQJ height significantly increased by 3.9 % following 20 Hz WBV, although there was no significant change in CMJ height. Lack of: standardisation for the semi squat posture; volume-matched controls; and it is unclear if blinding to the WBV frequencies used, limit the findings. In support of Cardinale & Lim (2003a), Da Silva-Grigoletto *et al.* (2006) also found 20 Hz increased SQJ height by a similar amount (3.7 %). A frequency of 30 Hz also significantly increased SQJ height.

In contrast to Cardinale & Lim (2003a), Da Silva-Grigoletto and co-workers reported 40 Hz significantly decreased SQJ height. However, one important WBV protocol difference is present. Da Silva-Grigoletto *et al.* (2006) used six repetitions of 60 s WBV, not five, and also had twice as long recovery periods. Da Silva-Grigoletto *et al.* (2006) found a significant increase in CMJ height following 30 Hz NEMES WBV. It seems that there is not one WBV frequency which improves jumping and squatting performances (Da Silva-Grigoletto et al., 2006). The posture adopted during the 60 s WBV was not given therefore, it is unknown if participants were in the same semi-squat posture as used by Cardinale & Lim (2003a). It is also unknown if participants were blinded and what footwear participants used during WBV, thus any potential dampening effect is undetermined.

The Kuntotary platform used by Torvinen *et al.* (2002c) produces a similar vertical oscillating stimulus to the NEMES and found no significant effect on isometric strength or CMJ performance. However, the D_{PTP} was 2 mm not 4 mm as used by the previous studies (Cardinale & Lim, 2003a; Da Silva-Grigoletto *et al.*, 2006). Therefore, it seems inappropriate to make direct comparisons, especially as the duration of the WBV exposure was four minutes continuous not 60 s repetitions.

Bullock *et al.* (2009) and Bullock *et al.* (2008) found no significant difference in jumping or sprinting performance post NEMES WBV in international skeleton athletes. It was suggested that the well-developed muscle tendon complex in high level athletes may reduce any potential improvement post-WBV (Bullock et al., 2008). It could be argued that 30 Hz and three repetitions of 60 s WBV may not have been a sufficient enough stimulus for these high level participants (Bullock et al., 2008). This is in conflict with Luo *et al.* (2005a). Bullock *et al.* (2008). did not utilise sport specific sprint tests but Bullock *et al.* (2009) did; still reporting no WBV response in sprint performance.

Elite athletes may have different musculotendinous characteristics leading to a mechanical type buffer which may dampen and reduce WBV transmission (Bullock et al., 2009). This is especially relevant as participants wore shoes during WBV exposure, known to influence the response to WBV (Marin et al., 2009). Therefore elite athletes may require higher WBV magnitudes, supporting the limitations of Bullock *et al.* (2008) as discussed previously. Finally, the timing of sprints (10 minutes post WBV) may be an issue as explosive performance responses have been reported to peak within 1 minute post WBV and significantly decrease up to 10 minutes post WBV (Bedient et al., 2009).

A detrimental response of WBV on IMVC in athletes has been reported (Jordon et al., 2010). The authors suggested that the acute WBV parameters used did not potentiate performance although, utilising isometric measures of performance instead of dynamic explosive performances may limit the findings. In addition, both WBV and control interventions resulted in reduced performance, suggesting a fatiguing exercise protocol. In conclusion, acute NEMES WBV may increase certain jump performances in general populations, although in trained individuals there appears little change in explosive performance or even a reduction following acute WBV.

2.4.2.4 Other whole body vibration platform types

Those studies which utilised other types of WBV platforms including Pneu-Vibe in trained athletes are summarised in Table 2.5. This includes Pneu-Vibe plus Med-Vibe platform types in trained participants; and iTONIC and Pneu-Vibe platform types utilising general populations.

Table 2.5: Summary of the acute effects of whole body vibration delivered by other WBV platform types on strength and power parameters of trained athletes. WBV, whole body vibration; Age, mean age in years; Hz, frequency; mm, peak to peak displacement; s, seconds; min, minute; m, metre; ↑ ↓, increase and decrease at significance level ($P < 0.05$); SQJ, squat jump; CMJ, countermovement jump; NS, non-significant; rec, recovery; IMVC, isometric maximal voluntary contraction; 1RM, one repetition maximum; NS, non-significant.

Author	**Participants**	**WBV device**	**Vibration characteristics**	**Outcome measures**	**Control group**	**Effects**
Dabbs *et al.* *(*2010)	n = 11 Age: 18 National softball players	Med-Vibe	25 Hz 13 mm 1 x 30 s Standing in normal softball hitting position	Bat swing speed	WBV group with additional 5 maximum bat swing 5 maximum bat swing only	NS difference between interventions
Guggenheimer *et al.* *(*2009)	n = 14 Age: 21 National level athletes	Pneu-Vibe	30, 40 and 50 Hz 1-2 mm 4 high knee repetitions lasting approximately 5 s	10, 20 and 40 m sprint times	Volume-matched control group	NS difference in sprint times across all WBV frequencies versus control
Rhea & Kenn (2009)	n = 16 Age: 23 1 year of resistance and plyometric training Unknown footwear	iTONIC	35 Hz 4 mm 1 x 30 s of unloaded squatting in between 3 x 75 % 1RM squats	Peak power during concentric phase of squat with 75 % 1RM load	Control sat resting in sitting position for 3min	WBV ↑ power by 5.2 % during a 75 % 1RM loaded squat
Ronnestad *(*2009)	n = 8 Age: 25 Resistance trained Shoes worn	Pneu-Vibe	20, 35, 50 Hz 3mm during loaded SQJ and CMJ repetitions performed at 90 ° knee flexion	SQJ and CMJ power via linear encoder	Volume-matched control group	20 Hz NS difference in SQJ or CMJ power 35 Hz NS difference in SQJ or CMJ power 50 Hz ↑ SQJ power 50 Hz NS difference in CMJ power

Author	Participants	WBV device	Vibration characteristics	Outcome measures	Control group	Effects
Ronnestad & Ellefsen (2011)	n = 9 Age: 23 Amateur soccer players Unknown footwear	Pneu-Vibe	Interventions: 1. 30 Hz 3 mm 2. 50 Hz 3 mm 1 x 15 squats during WBV	40 m sprint performance	Volume match control	50 Hz ↓ sprint time versus control NS change following 30 Hz versus control
Ronnestad *et al.* (2012)	n = 12 Age: 24 National power lifters Unknown footwear	Pneu-Vibe	50 Hz 3 mm during 2 x 3 65 kg SQJ & 2 x 3 100 kg SQJ repetitions	SQJ 65 kg and 100 kg peak power	Volume-matched control group	Peak power during both SQJ 65 kg and 100 kg ↑ during WBV versus control

Rhea & Kenn (2009) utilised a commercially available platform (iTONIC), and reported an increase in loaded squat peak power (5.2 %). However, the WBV stimulus was given during squatting and no volume-matched controls existed. Ronnestad & Ellefsen (2011) did utilise a volume-matched control involving an identical protocol without WBV, comparing both 30 and 50 Hz WBV frequencies. The 50 Hz frequency increased sprint performance, however no EMG measures were collected, nor any acceleration data to verify WBV frequencies as the authors admitted a ± 2 Hz manufacturer's validation. For example at 30 Hz, peak acceleration would be 71 $m.s^{-2}$ (Rauch et al., 2010). This variation would result in a peak acceleration of 62 to 81 $m.s^{-2}$. Conflicting the increase in sprint performance following 50 Hz WBV was Guggenheimer *et al.* (2009); who reported no WBV response on sprint performance. Although, the protocol involved exercising during WBV, potentially limiting the exposure to WBV as the effect of exercise during WBV exposure on the WBV magnitude is relatively unknown (see section 2.4.2.2).

Trained athletes increased SQJ power but not CMJ power following 50 Hz Pneu-Vibe WBV (Ronnestad, 2009; Ronnestad *et al.*, 2012). Limiting these findings somewhat are: unclear methodology regarding participants blinding and the use of footwear with a potential dampening effect. In addition the use of loaded SQJ and CMJ performances totalled up to 210 kg weight onto the WBV platform. The effect of this additional load on the WBV output is unknown (see chapter 7). The authors did comment that D_{PTP} was maintained even with a 300 kg load placed on the platform, although, this was never quantified and no acceleration data was presented.

Dabbs *et al.* (2010) found no significant difference following Med-Vibe WBV. The performance measure was bat swing speed and it is perhaps not surprising that WBV exposed to the lower limbs may not significantly influence an upper body performance measure. Finally, Kelly *et al.* (2010) reported Med-Vibe WBV was equally effective as cycling in terms of warm up. But this study was excluded on methodological grounds such as: no participant physical characteristics reported; the hip or knee angle adopted during WBV was not reported; and a lack of a volume matched control, comparing an isometric posture with dynamic cycling.

2.4.3 Summary of the responses to acute whole body vibration

The EMG response either during or following acute WBV appears to be higher than without WBV, although evidence remains conflicting. There is emerging evidence to suggest that there may be an individualised response of EMG to certain WBV parameters (e.g. frequency) and this warrants further research. The fact that higher EMG response during WBV has been reported still begs the question whether higher EMG magnitudes necessarily equates to improved performance? In trained individuals there are only a limited number of studies investigating these direct links between EMG response and performance measures.

Strength and power responses to acute WBV are generally beneficial, especially in jump performance measures. However, if the literature is divided by WBV platform type, Powerplate and NEMES platforms have either limited or mixed evidence regarding positive strength and power responses. When focussed on trained participants, the evidence for positive responses is even smaller, highlighting the limited literature available.

Overall, the evidence for acute WBV responses in strength and power is somewhat restricted due to: the lack of volume-matched control interventions/groups; performing exercise protocols during WBV; the use of un-quantified footwear and thus potential dampening of WBV stimuli; posture and exposure time during WBV protocols. A reason to explain the reduced effect within trained individuals may be already well developed neuromuscular systems and well adapted musculotendinous complexes.

2.5 THE POTENTIAL HARMFUL EFFECTS OF VIBRATION

As mentioned in section 1.1, there has been substantial research investigating the potential side effects of exposure to vibration. It is well established that exposure to large amounts of vibration or chronic exposure to vibration can result in negative effects, such as: hand-arm vibration syndrome and lower back pain (Jordan et al., 2005). Chronic WBV through occupational exposure has been reported to have several side effects, such as: disturbances of the visual and vestibular systems; and spinal vertebrae damage (Griffin, 1990; Abercromby *et al.*, 2007b). There are, to date, no standards established quantifying safe exposure to vibration as a training stimulus delivered via WBV (Jordan et al., 2005). As well as awareness of the negative effects, others have highlighted the level of care required to ensure safe protocols (Cardinale & Wakeling, 2005).

There is general agreement that WBV exposure as a training intervention is low in magnitude compared to that received through occupation (Dolny & Reyes, 2008). Nevertheless, caution is required as WBV has the capacity to expose an athlete to stimulus parameters that could cause injury (Jordan et al., 2005). In addition WBV has no appropriate standards which have been established, standardised or validated in the use of WBV for both training or therapeutic purposes (Prisby et al., 2008).

Reassuringly, a review (Wilcock et al., 2009) reported few side effects across the literature following WBV. One participant of a reviewed one hundred and sixty-six participants reported developing shin pain post WBV exposure and withdrew from the study. Contradicting this is the review by Dolny & Reyes (2008) who reported common WBV side effects of: erythema; itching; and oedema of the legs. How common these were was never quantified and all symptoms were reported as short lived, easing within a couple of minutes.

High transmissions of vibration to the head and torso have been recommended to be avoided to reduce the risk of negative effects of vibration stimuli on the visual and vestibular systems; in addition to the spinal vertebrae (Mester *et al.*, 2006; Dolny & Reyes, 2008). The magnitude of transmission can be assessed by accelerometers placed both on the WBV platform to monitor WBV output; but also placed on the user to monitor body segment accelerations (Rauch et al., 2010). As previously mentioned, posture especially knee angle can inversely influence transmission of acceleration to the head and torso (Abercromby et al., 2007b); leading to the recommendation that low knee flexion angles (0 – 20 º) should be avoided (Mester et al., 2006). This is supported as acceleration transmission to the head was minimised by up to 55.6 % during 26-30 º knee flexion squatting compared to 10-15 º knee flexion (Abercromby et al., 2007b). Lying or sitting on the vibration platform should also be avoided due to increased acceleration transmissions to the head and torso associated with those postures (Cardinale & Erskine, 2008). Although during this thesis there has been anecdotal evidence that some sporting backgrounds may specifically utilise this type of training for high vibration environments, e.g. motor sport driving.

A single case study (Monteleone et al., 2007) presented a 40 year old female amateur runner who was in good health with asymptomatic nephrolithiasis (kidney stones) and completed a single WBV session (30 Hz, 9 mm, 10 x 30 s exposure). It was reported that 12 hours post WBV exposure sudden pain onset occurred and the participant required medical and pharmacological intervention for nephrolithiasis. Clearly the past medical history is relevant in this single case study. It was recommended that suitable pre WBV investigations were completed in those populations at risk of renal calculi (Monteleone et al., 2007). Until very recently, nephrolithiasis was not included in contra-indication lists from some WBV manufacturers. This adds support to the recommendation that pre WBV screening is important to avoid exacerbating injuries (Jordan et al., 2005); and further investigation into all contra-indications has been recommended (Prisby et al., 2008). The available lists of contra-indications are supplied by manufacturers (see below) but there is a limited scientific research into this area.

Contra-indications to WBV:
Recent fractures or injuries which effected physical activity
Pregnancy
Acute thrombosis conditions
Cardio-vascular disease
Fresh wounds resulting from an operation or surgical intervention
Hip and knee joint replacements

Acute hernia, discopathy, spondylolysis
Diabetes
Known neurological conditions
Epilepsy
Heavy migraine
Fitted Pacemaker
Tumours
Symptomatic gallstones or kidney stones or asymptomatic history
Wearing recently fitted i.u. coils, metal pins, bolts or plates

Taken from "Galileo contraindications," (2007). Retrieved 12th May 2009, from http://www.galileouk.co.uk/contra.html.

A second case study (Franchignoni et al., 2012) highlighted a middle aged national level steeplechaser reporting haematuria symptoms (blood present in urine). The athlete appeared symptomatic with macroscopic (visible) haematuria following WBV sessions lasting 24 hours. A full medical assessment revealed microscopic haematuria at 40 hours post WBV which was clear 2 days later on re-test. The patient was advised to stop WBV use and on 1 year follow up remained symptom free. There appeared to be a temporal link between WBV use and symptoms, suggesting the most probable cause was repeated impact of the bladder wall (Franchignoni et al., 2012). Whether the mechanism of this is related to WBV is yet undetermined as haematuria can have no basis for symptoms (as in 61 % of cases) (Khadra et al., 2000). Sudden haematuria can be due to different causes, duration related (foot strike haemolysis) or intensity related (muscle tissue damage) (Franchignoni et al., 2012). The patient completed moderate intensity exercise the day before the onset of symptoms; however, it is unknown if it was steeplechase running.

One of the few reviews to investigate the possible side effects of WBV was Mester *et al.* (2006). The authors recommended that populations with coronary disease and hypertension should not participate in WBV (Mester et al., 2006). The review also investigated the safety considerations of WBV. It was recommended that frequencies lower than 20 Hz should be avoided. This was based on avoiding the resonant frequency of the body (5-10 Hz). Recommendations were made that small D_{PTP} (2 – 4mm) and short WBV exposure time (20 – 60 s) should be used initially (Mester et al., 2006). However, it is important to note that these recommendations were made based solely on cardiovascular response to WBV and the associated safety considerations.

Guidelines have been published quantifying an estimated vibration dose value (eVDV) (ISO, 1997). The eVDV may be used to quantify the exposure of WBV and relates to potential health risks (Abercromby et al., 2007b). These guidelines stated vibration may be potentially harmful if eVDV exceeded a value of 17. The eVDV was calculated for a daily 10 minute exposure to WBV and was found to exceed 17 (Abercromby et al., 2007b). The measure of eVDV has not been applied to WBV training and thus highlights a limitation in the ISO standards. Therefore, applying the ISO guidelines may not be appropriate as these were published for chronic exposure often in a sitting posture and relate to exposure to WBV through the buttocks (Rittweger, 2010).

During WBV exposure as a training stimulus the participants are more often standing, affecting the transmission magnitude of acceleration to the upper body and head. In addition, the exposure of WBV in training is rarely a continuous exposure for a prolonged period of time (e.g. occupational vibration over 8 hours using a pneumatic drill or sitting operating construction machinery). Therefore the ISO standards may not take into account WBV in the training situation (Issurin, 2005).

In theory, the WBV parameters that have the least potential for harmful effects are those that expose the participant to low head acceleration and low eVDV (Abercromby et al., 2007b). Even though ISO standards for WBV as a training stimulus may not be appropriate, at present there are no other applicable standards to apply to WBV.

2.6 CONCLUSIONS

Across the WBV literature there are differences in not only vibration delivery systems (i.e. locally applied versus WBV); but also in the form of different WBV platform types. The nature of WBV stimuli (magnitude and direction of vibration oscillation) are likely to have different characteristics, which may explain the conflict across the WBV literature. For example, a frequency which may elicit the most beneficial response. It appears that 30 Hz may result in the highest EMG activity response, whereas 40 Hz may give the most beneficial performance response. There is evidence to suggest that the response to WBV frequency may be individualised and perhaps gender-dependent. Further research is required to investigate the presently limited literature regarding the effects of D_{PTP} and exposure time within WBV protocols. It appears that exposure times of longer than 4 minutes are detrimental to performance and perhaps a shorter repetition length may improve jump performance responses to WBV. In terms of posture whilst on the WBV platform, evidence suggests that placing muscles under a degree of stretch may elicit a more beneficial WBV response.

Interest into WBV stemmed from research into TVR responses characterised by muscle contractions following application of vibration to a tendon. Debate exists in the application of TVR to a WBV setting, and overall mechanisms to account for the WBV response are still speculative. WBV may elicit small but rapid changes in muscle length detected by muscle spindles through a stretch reflex response. Via these responses neuromuscular facilitation may account for a WBV effect which may involve increased synchronisation, recruitment and coordination of motor unit firing. This may lead to increases in neuromuscular performance. Exposure to the hyper-g nature of WBV stimuli may also explain performance improvements especially via chronic protocols due to myogenic adaptations. WBV may evoke a dampening mechanism in response to stimuli characteristics close to the natural resonating frequency of soft tissue (muscle tuning). However, the link with performance benefits remains elusive. PAP may be another mechanism to account for the WBV response as the stimulus could be classified as a pre-conditioning activity. Although, the effect of these PAP mechanisms on performance remains to be investigated.

The acute WBV responses on EMG in general populations and trained individuals tend to be higher than the equivalent without WBV, but conflict does exist. Small amounts of evidence suggest that jump performance (such as force and jump height) increase following acute WBV. This appears more for general populations than for trained individuals. However, the vast majority of literature lacks volume-matched controls and adequate control for placebo and role of expectancy effects.

WBV has the potential for negative and harmful side effects if used incorrectly. The large amount of current literature demonstrating this investigated occupational vibration, which has different parameters than the application of WBV in a training setting. For example, frequency, D_{PTP} and exposure time parameters. Nonetheless, safe and effective protocols need to be established. This is problematic as no standards are specially designed for WBV in a training setting. Contraindications to WBV exist but these need regular reviewing and updating. Reassuringly, there appears a low occurrence of reported side effects and these seem temporary and minor. Based on transmission patterns and research into occupational vibration, postures which increase vibration magnitudes in the torso and head should be avoided.

Before safe and effective WBV protocols can be established, research is needed into the characteristics of the WBV stimulus itself. There is not sufficient evidence regarding the influence of specific parameters on the response of WBV users. But also, the influence those WBV users may have on the stimulus parameters itself. For example, the influence of acute WBV frequency on EMG and performance responses; and the influence of BM on WBV magnitudes. Therefore, the overall aim of this thesis was to investigate the characteristics of WBV stimuli and the acute neuromuscular responses to such stimuli, in an attempt to provide rationale behind safe and effective WBV protocols. Secondary aims were to investigate whether an individualised response to WBV frequency existed in EMG activity. Also how this may link with performance responses and whether an expectancy effect plays a significant role. An additional aim was to investigate the influence of data analysis methods following acute WBV.

Chapter 3: General methods

The methodology completed in chapter 4 utilised a unique protocol and therefore will be discussed in detail, see section 4.2.

3.1 PARTICIPANTS

Participants for chapters 5, 6 and 7 were recruited from within the University of Edinburgh involving both students and staff, following an extensive recruitment process. This involved contacting several university sport teams, poster displays and recruitment attempts within undergraduate and postgraduate classes. Inclusion criteria common to all studies included: age 18 to 35 years old; male, healthy active individuals, free from disease or injury; and a minimum of 6 months history of resistance training. Rationale for the chosen age range was one of both convenience and to follow ethical approval. The research for chapters 5, 6 and 7 was approved by the University of Edinburgh, Moray House School of Education Research Ethics Sub-Committee. Verbal approval from the Sub-Committee was granted for chapter 5 and 6, written approval was given for chapter 7, (see chapter 3 appendix). Recruiting young and healthy participants was to avoid associated exclusion criteria which may be present in more elderly populations and those with illness.

The reason for selecting male-only participants was to standardise participants as there is evidence to suggest a gender-specific response of WBV on jump performance (Bazett-Jones et al., 2008b) and also jump performance technique (Laffaye & Choukou, 2010). The rationale for including 6 months history of resistance training was to ensure participants were experienced in completing; maximal jump tests (chapters 5 and 6) and loaded squatting (chapter 7). In addition, the response of WBV in trained athletes may be higher than in non-trained (Cochrane et al., 2004). Exclusion criteria were chosen to avoid any potential negative or harmful effects of WBV. These are listed subsequently:

Recent fractures (stopping participants resuming normal training)
Recent injuries (resulting in a reduction of normal training)
Pacemaker
Hernia
Recent surgery (stopping participants resuming normal training)
Fresh wounds
Acute deep vein thrombosis
Cardiovascular disease
Hip or knee joint replacements
Spinal medical conditions such as discopathy or spondylosis
Epilepsy
Known neurological conditions
Joint or bone implants such as internal or external fixation
Diabetes
Tumours
Gallstones or kidney stones, symptomatic or asymptomatic

Taken from "Galileo contraindications," (2007). Retrieved 12th May 2009, from http://www.galileouk.co.uk/contra.html.
(Monteleone *et al.*, 2007)

Once participants fulfilled inclusion criteria they were given information sheets and provided informed consent for chapters 5 to 7 (see chapter 3 appendix). Participants' height was recorded for all studies using a stadiometer (Seca, model 225, Hamburg, Germany). Participants' BM was recorded differently across studies, see individual study method sections. A summary of participants' physical characteristics is presented in Table 3.1.

Table 3.1: Physical characteristics of participants recruited to chapter 4, 5, 6 and 7 studies. N, participant number in each study; Mean (± SD) for age, height and weight; cm, centimetre; kg, kilogram. (Data recorded from participants in chapter 5 were used in chapter 6, therefore participants in both chapters are identical).

Study	N	Age (years)	Height (cm)	Weight (kg)
Chapter 4	28	26 ± 4	187 ± 8	104.2 ± 13.2
Chapter 5	7	23 ± 4	180 ± 6	82.6 ± 9.0
Chapter 6	7	23 ± 4	180 ± 6	82.6 ± 9.0
Chapter 7	10	26 ± 5	176 ± 6	73.4 ± 3.1

3.2 SURFACE ELECTROMYOGRAPHY

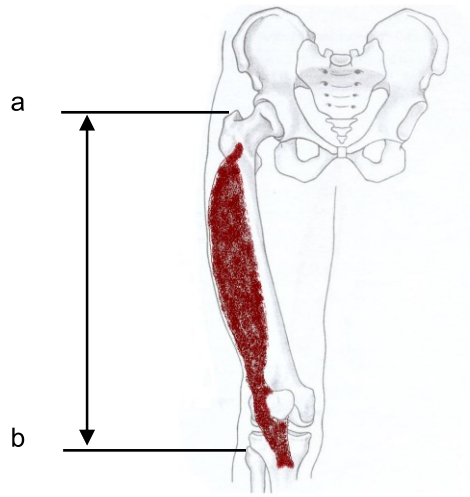
3.2.1 Procedure

Skin preparation for electrode placement followed identical procedures for all studies. That is, participants' skin was shaven, cleaned with alcohol and left to vaporise to ensure dry skin (Hermens et al., 2000). Common, standardised electrode placement procedures across all studies involved: electrode orientation parallel to muscle fibres; and utilising double-sided adhesive sensor interface to fix electrodes onto the skin (Hermens et al., 2000). In addition, cables were secured to minimise motion artifacts (Bartlett, 2007) and a reference electrode was attached to the left anterior superior iliac spine (Delsys Inc., 2008).

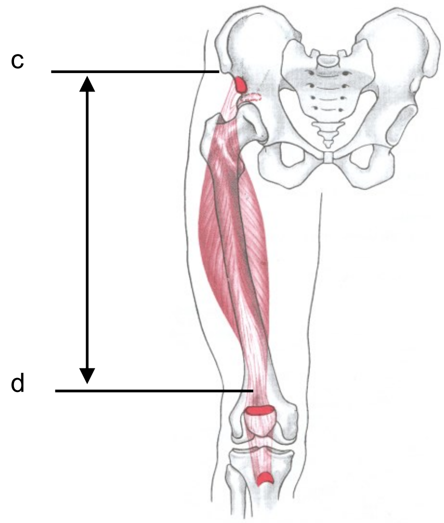
Electrode placement per muscle was study-specific and followed the principles of Bartlett (2007) and Hermens *et al.* (2000) such as: electrode placement over the mid-point of the muscle belly; in orientation parallel to the direction of muscle fibres. Chapter 5 involved repeat measures across multiple visits, therefore standardisation for each visit was ensured by both marking electrode locations onto participant's skin via permanent marker; and by recording electrode locations in relation to anatomical landmarks. For chapter 5, vastus lateralis, rectus femoris and biceps femoris muscles were recorded, for locations see Figure 3.1. For each muscle 50 % of the distance between the corresponding anatomical landmarks was chosen for the electrode placement site. This was to ensure standardisation and also that the electrode placement site was on the muscle belly.

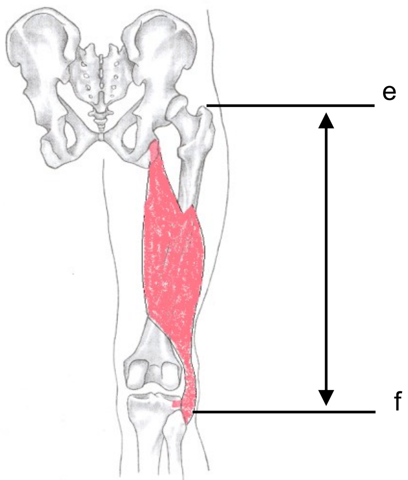
Figure 3.1: Electromyography electrode placements for vastus lateralis, rectus femoris and biceps femoris muscles (Thompson & Floyd, 1998; Bartlett, 2007).

Vastus lateralis: electrode placement between anatomical landmarks; most superior aspect of greater trochanter (a) and most superior aspect of fibular head (b).



Rectus femoris: electrode placement between anatomical landmarks; anterior superior iliac spine (c) and superior aspect of the patella (d).

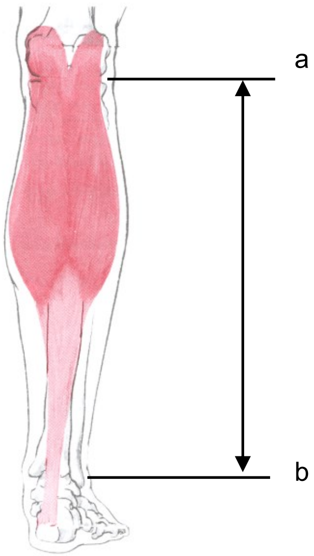




Bicep femoris: electrode placement between anatomical landmarks; most superior aspect of greater trochanter (e) and most superior aspect of fibular head (f).

As chapter 6 utilised the same EMG data, electrode placement positions were identical. Chapter 7 utilised both vastus lateralis and rectus femoris EMG data (Figure 3.1) but also recorded gastrocnemius, ¼ of the distance between most superior aspect of fibular head and lateral malleolus (Figure 3.2). For all studies a reference electrode placed on the left iliac crest was utilised. The use of percentile distances from anatomical landmarks to identify electrode placement has been demonstrated as reliable (Intraclass Correlation Coefficient > 0.8) during IMVC and jump tests (Fauth et al., 2010)

Figure 3.2: Electromyography electrode placements for gastrocnemius muscle (Thompson & Floyd, 1998; Bartlett, 2007).



Gastrocnemius: electrode placement ¼ the distance from anatomical landmark, 'a' (most superior aspect of fibular head) to 'b' (lateral malleolus).

3.2.2 Equipment

EMG data were recorded using rectangular shaped (20 x 35 x 5 mm, width, length and depth respectively) bipolar surface electrodes with: 1 x 10 mm 99.9 % Silver (Ag) conductor; an inter-electrode distance of 10 mm housed in a shielded polycarbonate plastic shell; and a typical common mode rejection ratio (CMRR) of 92 dB with a minimum CMRR of 84 dB (DE 2.1, Single Differential Surface EMG Sensor, Delsys Inc., Boston, USA). Reference electrodes were also used with a 5.1

cm diameter sensor (Dermatrode ®, American Imex, Irvine, USA). Electrodes were connected via an intermodule to a 16-channel, fixed shielded cabled EMG system (Bagnoli™ Main Amplifier Unit, Delsys Inc., Boston, USA) via an input module and then to a main amplifier unit. For all studies EMG data were recorded at sample rate of 1000 Hz, a 1000 gain. The sample rate of 1000 Hz was determined using the Nyquist theorem; stating that EMG signals recorded should be sampled at a rate twice the frequency of the highest signal harmonic (Merletti & Parker, 2004). A typically minimum sampling rate has been suggested as 1000 Hz (Payton & Bartlett, 2008).

A gain of 1000 was selected based on superficial muscle activities in an exercise application (Payton & Bartlett, 2008). Conversion from analogue to digital (A/D) was achieved via a 16 bit A/D system set at ± 5 V. This was to ensure small changes in EMG activity were detected based on typical EMG signal amplitudes (e.g. 4 – 5 mV) (Merletti & Parker, 2004; Payton & Bartlett, 2008). Via the A/D system a built in electric band pass filter (20 – 450 Hz) was utilised through EMG data acquisition software (EMG Acquisition, EMGworks 3.1, Delsys Inc., Boston, USA) onto a PC laptop. The electric band pass filter was chosen to eliminate electrical and tissue noise at electrode site (Cram & Kasman, 1998). EMG data was recorded during IMVC and during WBV for chapter 5 and 6; and during WBV only for chapter 7.

3.2.3 Data analysis

Once filtered and saved EMG data across all studies was analysed by software (EMG analysis, EMGworks 3.1, Delsys Inc., Boston, USA) via a PC laptop. For chapter 4 raw EMG data was rectified and smoothed and finally direct current removed (Spike 5 software, Cambridge Electronic Design 1401, 2005). Mean raw EMG data was then calculated before EMG_{rms} was derived (unknown width of time window, default setting within initial Spike software used). Fast Fourier Transform (FFT) calculations were then performed (window type: Hanning; window length: 0.82 s) (Payton & Bartlett, 2008).

Data analysis for chapter 5 of utilising the EMG system electric filter (band pass filter 20 – 450 Hz). For chapter 6, FFT were performed on raw EMG signal to present Power Density Spectra. The FFT were performed as follows: window type, Hanning; FFT length 2.048; window length, 0.125; and window overlap 0.0625 (Payton & Bartlett, 2008). Finally for chapter 7, a digital band pass filter was applied, see section 7.2.5.1. Following filter applications for chapters 5, 6 and 7 studies, EMG data was root mean squared utilising a window length of 0.125 and window overlap of 0.0625. The use of a relatively long duration of time window was due to restrictions within the Delsys software and it is acknowledged that longer widths may result in rapid changes of EMG activity going undetected (Payton & Bartlett, 2008). However, the use of an overlapping window offers a solution to lessen the impact of longer window durations (Payton & Bartlett, 2008).

3.3 WHOLE BODY VIBRATION

3.3.1 Procedure

Participants in all studies stood on the WBV platform barefoot for standardisation, as potential dampening effects of footwear could vary. In addition, participants exposed to WBV barefoot elicited higher EMG responses than when shoes were worn (Marin et al., 2009). Participants stood in the centre of the platform and were advised to distribute their weight predominantly onto the balls of their feet; with their heels remaining in contact with the platform. This was standardised as vibration transmission can be affected by posture and likely weight distribution (Kiiski et al., 2008). A 90 º knee flexion posture was adopted, aiming to place the quadriceps muscle group under a degree of stretch to maximise any potential WBV benefits (Cardinale & Lim, 2003b; Ritzmann *et al.*, 2012). In addition, adopting this posture and avoiding heel stance reduces transmission of vibration to the head and torso (Matsumoto & Griffin, 1998; Mellor & Hodges, 2006; Abercromby *et al.*, 2007b), minimising potential negative effects of WBV exposure (Rittweger, 2010).

Participants were instructed to use the handrails for support due to safety and for standardisation. For the control condition, participants performed an identical procedure, however, the platform was not switched on for the duration.

3.3.2 Equipment

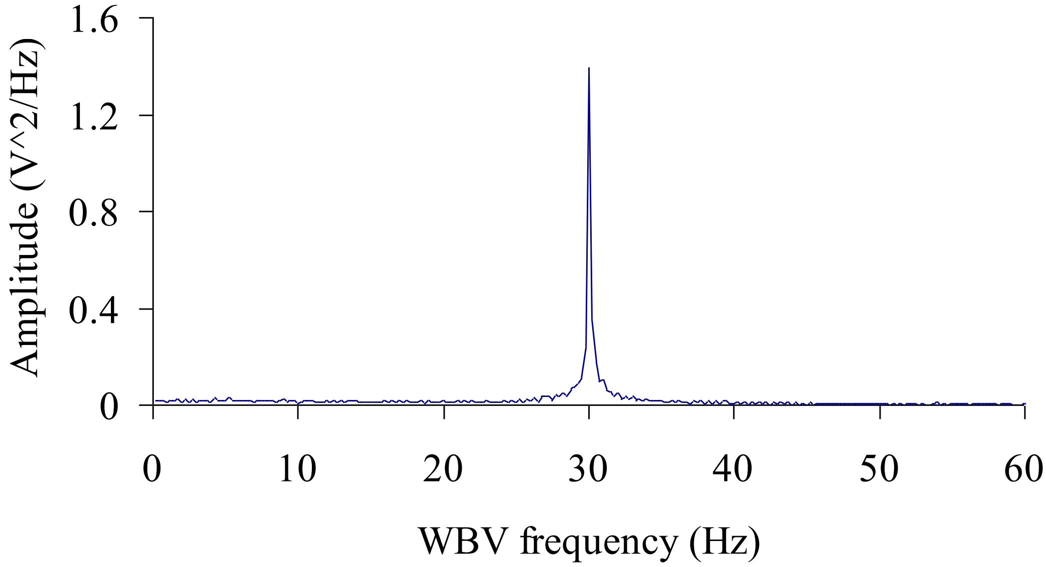
For chapters 5, 6 and 7 WBV was delivered via a synchronous vertical oscillating WBV platform (VM Master, VibraMachines Ltd., Ledborough, UK). Initial work presented in chapter 4 utilised a different WBV platform, as applied sport science support was given to a squad already using a pre-existing WBV platform (for methodology, see section 4.2). The WBV platform has pre-set frequencies of 30, 35, 40 and 50 Hz. Peak to peak displacement was set to low (3 mm). Peak acceleration for the four pre-set frequencies are calculated as follows (Rauch et al., 2010):

30 Hz:	$2 \times \pi^2 \times 30^2 \times 0.003$	$= 53.3\ ms^{-2}$	$= 5.4\ g$
35 Hz:	$2 \times \pi^2 \times 35^2 \times 0.003$	$= 72.5\ ms^{-2}$	$= 7.4\ g$
40 Hz:	$2 \times \pi^2 \times 40^2 \times 0.003$	$= 94.7\ ms^{-2}$	$= 9.7\ g$
50 Hz:	$2 \times \pi^2 \times 50^2 \times 0.003$	$= 148.0\ ms^{-2}$	$= 15.1\ g$

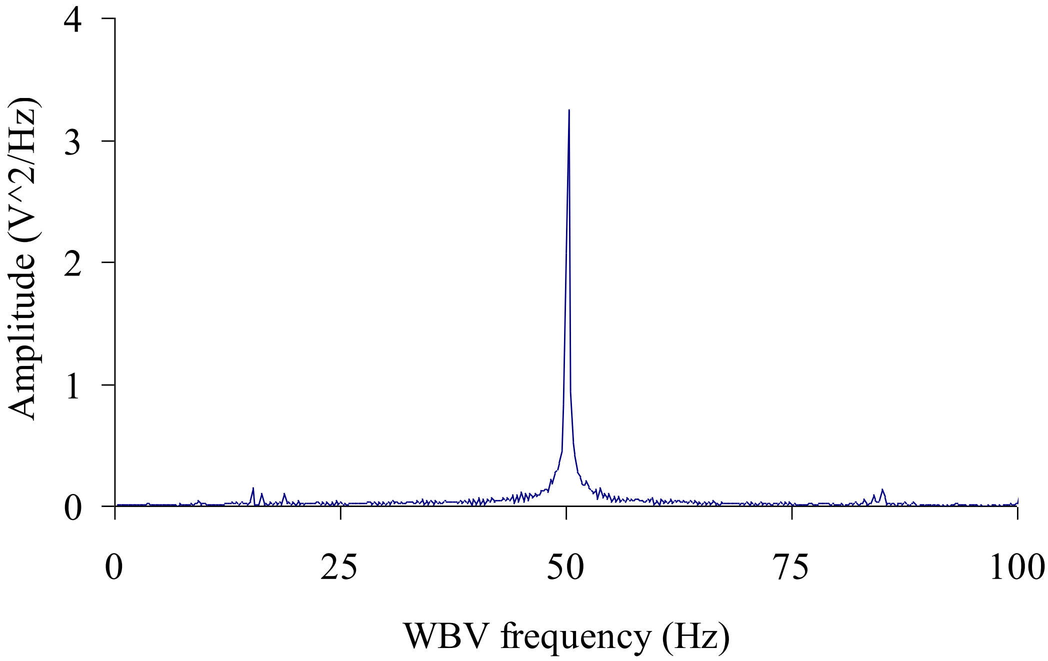
The accuracy of frequency output by the VM Master platform was analysed by means of FFT as in Rauch *et al.* (2010) (see Figure 3.3). At all frequencies (30, 35, 40 and 50 Hz) the true output was within ± 0.25 Hz suggesting good accuracy of frequency output.

Figure 3.3: Fast Fourier Transform analysis to calculate Power Spectra Density for two examples of signal recorded from VM Master platform at 30 and 50 Hz. Both main peaks indicate the actual WBV frequency is similar to that displayed by the platform.

30Hz



50Hz



3.4 STATISTICAL ANALYSIS

All statistical analyses were performed using the same software package (PASW Statistics version 17 for Windows, SPSS Inc., Chicago, USA). All data was subjected to data checks for normality and sphericity (Mauchly's test). If sphericity was violated then degrees of freedom were corrected using Greenhouse – Geisser estimates of sphericity quoting epsilon adjustment values (Field, 2005).

A common statistical analysis method utilised across all studies was analysis of variance (ANOVA) and therefore the basic procedure common to all ANOVA's performed will be covered to avoid repetition. Significance level was set at $p < 0.05$. Where a significant main effect *F*-value was determined, post hoc paired *t*-tests with Bonferroni correction were used to locate the significant differences. Where a significant interaction effect *F*-value was determined at higher level of ANOVA (e.g. three-way or two-way ANOVA); then a subsequent lower level of ANOVA was performed. At each level post hoc *t*-tests with Bonferroni correction were used to locate the significant differences. Each significant interaction effect was further explored down to one-way ANOVA level.

For all ANOVA tests, Partial Eta^2 values were calculated and expressed as effect sizes ($r = \sqrt{}$ Partial Eta^2): small effect sizes are considered $r = 0.1$; moderate effect sizes are considered $r = 0.3$; large are considered $r = 0.5$; very large are considered $r = 0.7$; and near perfect effect sizes are considered $r = 0.9$ (Hopkins, 2001).

Chapter 4: Study 1

4.1 INTRODUCTION

As outlined in section 2.2.3.1 the effect of WBV frequency on the neuromuscular response is both varied and conflicting. Fattorini *et al.* (2006) reported EMG response may be dependent on oscillation frequency. For example, 30 Hz has been shown to elicit the highest EMG response in elite volleyball players (Cardinale & Lim, 2003b). In terms of explosive performance it appears that 30 Hz also improves SQJ and CMJ performance (Da Silva-Grigoletto et al., 2006). Although, 20 Hz was reported to improve SQJ performance (Cardinale & Lim, 2003a) and 40 Hz to improve CMJ performance (Bazett-Jones *et al.*, 2008b; Turner *et al.*, 2011). It is worthwhile noting that these studies all utilised similar WBV platforms (NEMES and Powerplate), oscillating in a predominantly vertical direction. It appears that the effect of WBV frequencies on neuromuscular response is conflicting, and remains relatively unknown. This is especially true for well-trained participants, as the response to WBV may well be specific in these individuals compared to untrained participants (Luo et al., 2005a).

For well-trained participants and at an individual level it is unclear what the most effective WBV frequency would be to elicit a beneficial neuromuscular response. At present, in WBV literature, all bar two studies (Di Giminiani *et al.*, 2009; Di Giminiani *et al.*, 2010) have utilised a pre-determined WBV frequency applied across all participants. Even though previous work has suggested large inter-participant variability in EMG and explosive jump performance responses to WBV frequency exists (Da Silva-Grigoletto *et al.*, 2006; Di Giminiani *et al.*, 2009).

The work by Di Giminiani *et al.* (2009) was the first to assess individuals' EMG response to varying WBV frequency. Based on their findings, individuals were assigned the frequency of WBV which elicited the highest acute EMG response. This frequency was then utilised during a chronic WBV programme. A significantly greater SQJ performance improvement was found following an individualised programme versus a pre-determined WBV frequency for all participants (11 versus 3 % increases respectively). As already mentioned in section 2.3 there may be several reasons to explain this, such as individual differences in: muscle spindle properties; mechanoreceptor quality and location; visoelastic properties of the muscle; and muscle fibre percentage (Cardinale & Lim, 2003a).

The findings by Di Giminiani *et al.* (2009) are limited to physically active participants. Literature utilising well-trained participants is lacking. In addition, the study investigated the chronic effects of WBV. As such there is very little literature investigating the acute EMG response in well-trained athletes to different WBV frequencies. An opportunity arose to investigate WBV in elite athletes as an invitation was offered to assist in applied sport science work for an international rugby squad during a training camp. In return for investigating and supplying management staff with player-specific reports of the acute responses to WBV, the data could form an initial study for this thesis and an opportunity to learn how to collect and analyse such EMG data. This study differs from the work by Cardinale & Lim (2003b) in the participants' sporting backgrounds as rugby and volleyball players are likely to differ in: anthropometric and strength characteristics; typical training programme design; and likely training adaptations. These would be more explosive in nature in volleyball players, compared to more strength related in rugby players.

Therefore, the aim of this initial study was to determine what frequency of WBV elicits the highest EMG response in elite rugby players. The hypothesis was that 30Hz WBV would elicit higher EMG responses than other WBV frequencies, as previously demonstrated in elite volleyball participants (Cardinale & Lim, 2003b).

4.2 METHODS

4.2.1 Participants

In total 28 international level rugby players took part (age, 26 ± 4 years; height, 187 ± 8 cm; mass, 104.2 ± 13.2 kg). As data collection occurred as part of an international training camp, participants gave informed verbal consent after explanation of the experimental procedure. All participants had an extensive history of resistance training, associated with several years of rugby experience at both professional club and international levels.

4.2.2 Whole body vibration

Participants were exposed to WBV via sinusoidal tri-planar oscillations on a WBV platform (Powerplate® Pro5™, Powerplate International, The Netherlands). This type of platform has been described as tri-planar (Powerplate, 2013), but also as vertical (Abercromby et al., 2007a). Each participant stood barefoot at a visually monitored 100 º knee flexion posture. This was an attempt to place the vastus lateralis muscle under a degree of stretch to maximise potential WBV effects (Eklund & Hagbarth, 1966; Cardinale & Lim, 2003b). Participants were instructed to cross their arms over their chests to standardise upper movement during WBV exposure. Each participant was exposed to 15 s of WBV stimulus at 30, 35, 40 or 50 Hz during a constant 4 mm peak to peak displacement. The order of frequencies was randomised and blinded from participants. Each participant was given 30 s recovery between each bout of WBV, in which they stood upright on the platform. This recovery period was governed by time issues due to the nature of data collection during an international training camp.

4.2.3 Electromyography

Right vastus lateralis EMG signal was collected, using Ag surface disc electrodes (5mm recording surface and inter-electrode distance of 20 mm). Electrodes were placed approximately two thirds of the length between anterior spina iliac superior and lateral side of the patella (Eckhardt et al., 2011). The same electrode placement was maintained throughout the protocol by securing electrodes using adhesive tape. The vastus lateralis muscle was not difficult to identify in these participants due to high muscle definition. A third ‘earth’ disc electrode was utilised.

The raw EMG signal was recorded using a multi unit surface recorder (Bioelectronics Unit, Glasgow University, x 1000 gain, 20 – 1000 Hz inbuilt band pass filter). EMG signal was recorded during the middle 10 s of WBV stimulus. This was to allow for the WBV platform to reach constant and stable output after it was switched on, and to avoid including any deceleration of output when the platform was switched off.

4.2.4 Data analysis

Raw EMG data was rectified and smoothed and finally direct current removed (Spike 5 software, Cambridge Electronic Design 1401, 2005). Mean raw EMG data was then calculated before EMG_{rms} was derived. Upon visual inspection a total of 4 participants’ EMG data were unusable due to technical faults. These were excluded from the subsequent results, resulting in n = 24. The included data was then subject to FFT calculations as discussed in section 3.2.3.

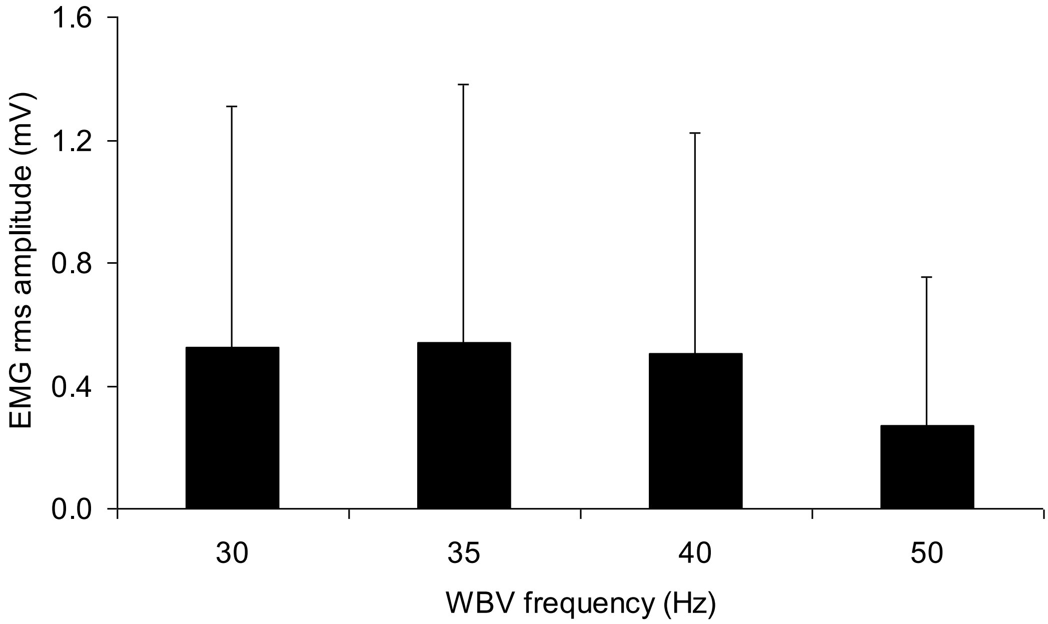
4.2.5 Statistical analysis

Data checks for normality and sphericity were completed (see section 3.4). The effect of WBV frequency on EMG response was analysed by means of a 1-way repeated measures ANOVA (frequency [30, 35, 40, 50 Hz]). Mean EMG_{rms} data was subjected to within participant analysis, consisting of mean and standard deviation (SD) calculations. Within participant variance was calculated as the mean of SD^2 of each WBV frequency. Within participant percentage coefficient of variation (% CV) was also calculated (Hopkins, 2000).

4.3 RESULTS

All participants completed the protocol without any reported side-effects and anecdotally reported the short protocol as enjoyable. No significant main effect for frequency was found (F [1.97,45.30] = 2.09 p = 0.14) with small effect size (r = 0.28). The mean values (+ SD) EMG_{rms} amplitude during four whole body vibration (WBV) frequencies are presented in Figure 4.1.

Figure 4.1: Mean + SD EMG_{rms} amplitude of vastus lateralis during 10 s of WBV at four different frequencies, no significant main effect for frequency was found.



Large within participant SD's were found, (± 0.62 mV). Mean % CV for within participant data was high with an equally high SD (47.5 ± 36.7 mV), suggesting large inter-participant variability. Figure 4.2 represents the EMG_{rms} amplitude of the WBV frequencies across participants. Figure 4.3 represents a typical sample of FFT calculations performed across participants during different WBV frequencies. Apparent motion artifacts are evident at frequencies corresponding to the fundamental WBV frequency chosen during the particular EMG recording.

Figure 4.2: EMG_{rms} amplitude per WBV frequency (30, 35, 40 and 50 Hz) across participants.

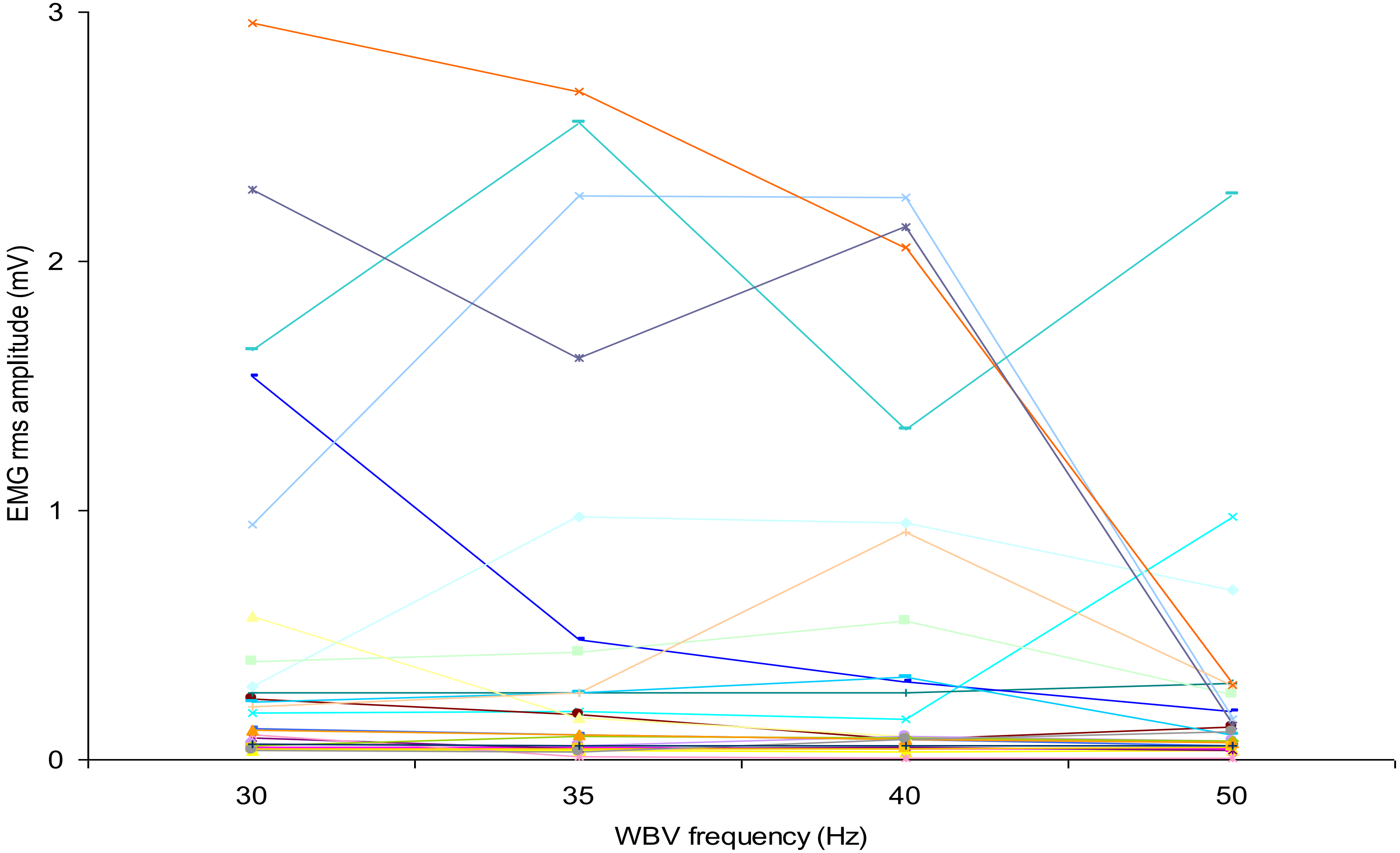
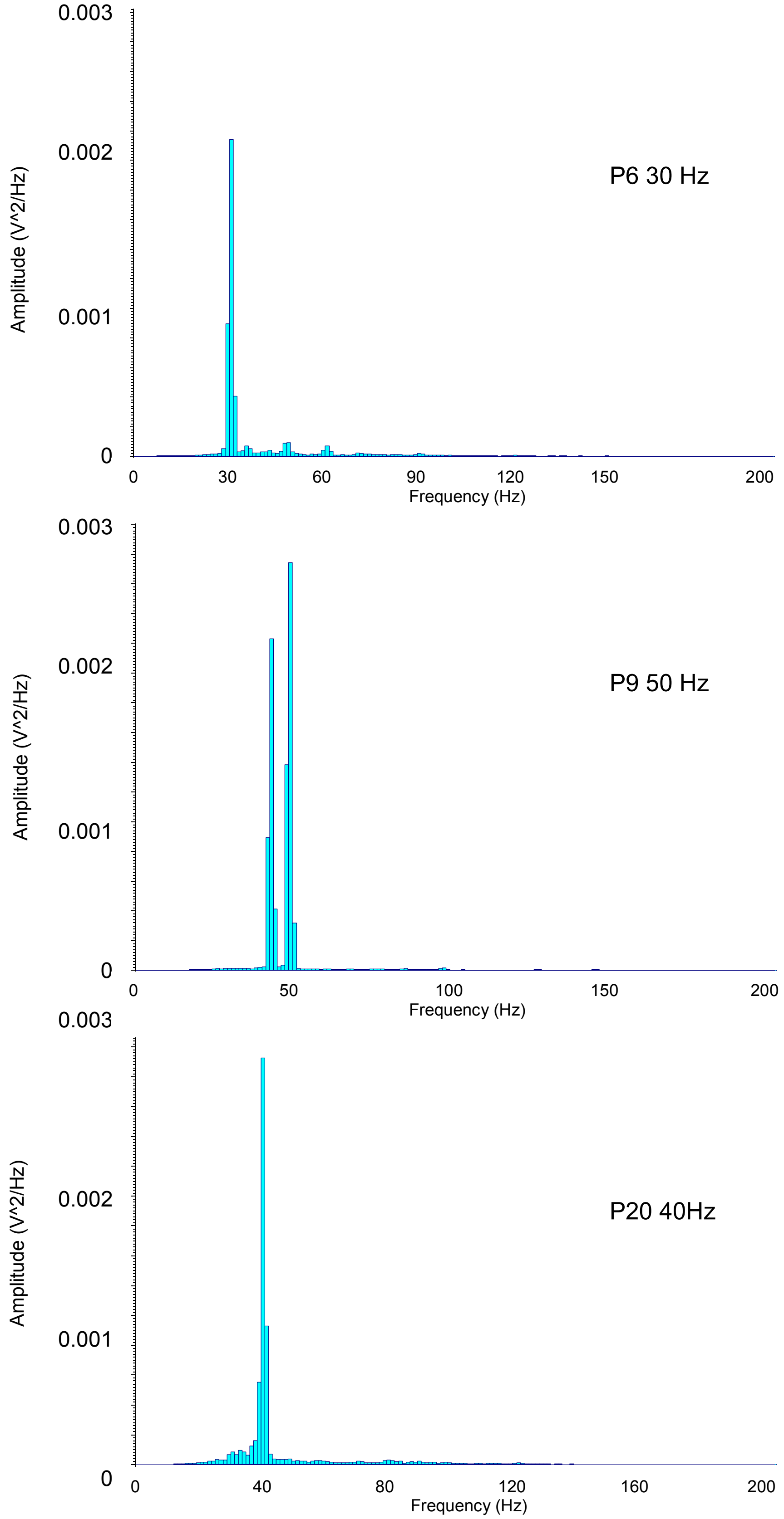


Figure 4.3: Fast Fourier Transform analysis of sample data of EMG signal showing apparent motion artifacts at the corresponding fundamental WBV frequencies.



4.4 DISCUSSION

The aim of the initial study was to determine what frequency of WBV elicited the highest EMG response in elite rugby players. The results demonstrate that across participants as a group, no one single WBV frequency elicited a significantly greater response in EMG than another WBV frequency, during 100 º knee flexion squat posture. Therefore, the hypothesis that 30 Hz would elicit the highest EMG response across these participants as a group can be rejected. The results from this study also suggest high inter-individual variability.

The findings of this study conflict with those of Cardinale & Lim (2003b), who reported 30 Hz was associated with the highest EMG response. Given the number of similarities between this present study and Cardinale and Lim (2003b) (elite athletes; WBV frequencies; posture adopted during WBV exposure; and finally muscle group EMG recordings); it is perhaps surprising the findings conflict. However, there are a number of differences that may explain the conflicting results.

Firstly, a difference in WBV platform type between (vertical versus tri-planar) may explain the conflict, due to different vibration oscillating directions as previously literature has suggested an oscillating specific response in vertical versus tri-planar vibration stimulus types (Bullock *et al.* (2008) *cf.* Adams *et al.* (2009)), see section 2.2.2. However, the response was noted in CMJ performance and not in EMG response as measured by this present study. Previous literature has investigated the response of EMG suggesting varying oscillating stimuli has a significant impact (Abercromby *et al.*, 2007a, b); albeit between tri-planar versus rotational. How EMG response may vary between tri-planar and vertical oscillating stimuli remains unknown. It appears plausible that the difference in platform type may explain the conflicting findings between this present study and Cardinale & Lim (2003b). On visual inspection there appears similar variation (as characterised by SD) in EMG responses in the work by Cardinale & Lim (2003b) and this initial study.

Secondly, difference in D_{PTP} between Cardinale & Lim (2003b) and the present study is also important (10 versus 4 mm respectively). Although, previous literature (Bedient *et al.*, 2009; Armstrong *et al.*, 2010; Gerodimos *et al.*, 2010) has suggested minimal impact of varying D_{PTP} on performance (see section 2.2.3.2); this was restricted to jump performance and to this author's knowledge there a scarcity of research investigating the influence of varying D_{PTP} on EMG response. Nevertheless, the difference between 10 and 4 mm D_{PTP} may be significant in explaining the conflicting findings. It would seem reasonable that if the direction of oscillation influences EMG activity (Abercromby et al., 2007a); then the magnitude of that oscillation would likely influence EMG response. For example, at 30 Hz, 4 and 10 mm would equate to peak acceleration values of 71.1 and 177.7 $m.s^{-2}$ respectively (Rauch et al., 2010).

Thirdly, differences between this present study and Cardinale & Lim (2003b) in WBV exposure time (15 s versus 60 s respectively). Even though previous literature has suggested a minimal effect of WBV exposure time on performance (Adams *et al.*, 2009; Da Silva-Grigoletto *et al.*, 2011); this has focussed upon the effect on jump performance and not on EMG response. It could be speculated that a longer exposure time to WBV stimulus may have given different EMG responses to varying WBV frequencies. This aspect of the present study was somewhat out of the experimenter's control as data collection was undertaken during a training camp. Due to the squad numbers and time restraints of four sets of WBV frequencies, only 15 s repetitions could be utilised.

Fourthly, differences in sport-specificity may explain the conflict. Although both participant groups were elite in standard; the present study recruited international rugby players whereas Cardinale & Lim (2003b) recruited volleyball players (mean BM 104.2 and 75.1 kg respectively). The influence of this difference in BM on the respective WBV platforms is unknown (but will be discussed in chapter 7). Due to difference in training types, (volleyball players and plyometric explosive type training versus rugby players, and more typical strength type training); variation in muscle fibre characteristics of these two participant groups, may result in different

EMG responses to WBV. Overall strength is likely to be different between these two participant groups which may explain the differing WBV responses. For those participants involved in predominantly strength type training PAP responses to WBV may be more pronounced (Hamada et al., 2000). Whereas those participants involved in plyometric training are likely to have a greater explosive capacity and thus respond in EMG activity differently to WBV.

The present study findings also conflict with those of Di Giminiani *et al.* (2010) who reported a mean frequency for all participants of 38.5 ± 9.14 Hz eliciting the highest EMG response. Comparing Di Giminiani *et al.* (2010) with the present study, differences in WBV platform types and D_{PTP} magnitudes exist. The use of a reported mean value by Di Giminiani *et al.* (2010) may be somewhat misleading. This was a mean analysis of each participant's iWBV frequency, and no other statistical analyses (such as ANOVA's) were completed. If a similar mean analysis was applied retrospectively to this present study data, the mean frequency would be 35.4 ± 6.7 Hz (a difference of 3 Hz). Even through this perhaps limited statistical approach, the average frequency which elicited the highest EMG response is lower than that Di Giminiani *et al.* (2010). Differences in training status of participants (sport science students versus elite rugby players) may explain the conflict. However, both this present study and Di Giminiani *et al.* (2010) reported large inter-participant variability. The findings reported here support that of Di Giminiani *et al.* (2010) that individual response to WBV frequency is evident.

To the author's knowledge this is the first study to investigate the EMG response to varying WBV frequencies in elite rugby players. The findings that individual responses may exist supports that of Di Giminiani *et al.* (2009). Even with a difference in WBV platform type, a similar individualised response was reported, in which the inter-participant EMG response varied considerably.

As discussed in Chapter 2, the TVR response is thought to be proportional to frequency (Eklund & Hagbarth, 1966). It would be reasonable to expect that 50 Hz may elicit the highest EMG response. This was found not to be the case in the present study; in fact, 50 Hz had the lowest mean EMG_{rms} amplitude response. However, original TVR research involved vibration stimuli applied directly to the muscle or tendon; not via WBV. The EMG response following WBV is likely to involve other neural pathways (refer to section 2.3.3) (Cardinale & Bosco, 2003; Pollock *et al.*, 2012). The mechanisms underlying the TVR response are thought to be both spinal and cortical in nature (Cardinale & Bosco, 2003). Both the transmission of vibration through the body and these spinal and other neural pathways are likely to individual specific.

The inclusion of a spinal stretch reflex loop, muscle spindles are included, which detect the small rapid changes in muscle length during exposure to WBV. It is thought that vibration stimulus increases synchronised firing of muscle spindles (Jordan et al., 2005); which may lead to overall higher synchronisation of muscle activity, accounting for greater EMG activity (neuromuscular facilitation) (Cardinale & Lim, 2003b) (see section 2.3.4). Muscle spindle sensitivity may increase (Ronnestad, 2004) via an increase in α motor neurone flow (Cardinale & Bosco, 2003). Through this increased spindle sensitivity a reduction in firing thresholds or previously inactive motor units may occur (Pollock et al., 2012). Increased EMG activity during WBV stimulus may also be explained by altered recruitment patterns via a dampening mechanism to the WBV stimulus (Mischi et al., 2010). These several mechanisms may explain increased EMG activity during WBV; however, they also give possible reasons behind an individualised response to WBV characteristics (specifically frequency).

This follows works of Di Giminiani *et al.* (2009; 2010) who both reported inter-participant variability. Large individual responses were suggested as differences in EMG activity of up to 74 % were dependent on which WBV frequency was utilised (Di Giminiani et al., 2009). Inter-participant EMG activity differences of greater than 74 % were found, following WBV frequencies of 30, 35, 40 and 50 Hz; possibly signifying greater differences in sensitivity to WBV frequency (Di Giminiani *et al.*, 2010). The greater inter-participant differences in EMG activity between frequencies were shown in elite trained athletes versus active but untrained individuals. It could be speculated that the more trained an individual is, the higher the potential variability and individualised response to WBV frequency may be. This may be due to increased fast twitch muscle fibre characteristics and motor unit recruitment; however no direct link between muscle fibre characteristics and individualised WBV responses has been investigated as yet.

Armstrong *et al.* (2008) investigated the response of a neuromuscular excitability measure (H-reflex) following WBV; reporting high variability across a range of participants training states. These training levels included sedentary to university athletes and the individualised response in afferent excitability appeared associated was not accounted for by gender (Armstrong et al., 2008). However, these specific variables did not appear to be statistically analysed as a direct research aim. The authors did offer an alternative explanation that muscle fibre characteristics may play a role in a potential individualised response to WBV. These characteristics may be largely determined by genetics but also by age and training activity (Sale, 2002). Individual differences in: muscle spindle band width properties; mechanoreceptors and proprioceptors quantity and location; and finally percentage of type II fibres, may all contribute towards the ability to dampen the WBV stimulus itself, but also the varying frequencies (Cardinale & Lim, 2003a). It seems reasonable to suggest that this highly trained elite athlete population in the present study would likely differ in most, if not all, of these categories. This may explain the greater difference in EMG activity dependent on WBV frequency in this present study versus previous literature which utilised relatively untrained participants.

If a potential individualised response to WBV stimulus exists, it could be argued that a similar individualised responses to WBV in performance measures would be evident. The vast majority of literature has utilised one single WBV frequency across all participants. This may explain non-significant differences to WBV reported with high variability influencing overall findings (Lamont et al., 2009). Individual responses to WBV may be related to dampening abilities as individual variability has been reported in accelerations measured at higher body segments at: the shank and thigh levels (Cook et al., 2011); but also at head level (Abercromby et al., 2007b). This individual ability to dampen WBV stimuli will be discussed in chapter 7.

The lack of muscle biopsy measures in this present study highlights a limitation but also means the previous discussion points regarding reasons for an individualised response to WBV are speculative. Due to the highly trained elite status of participants and timing of data collection (pre international fixtures), the use of biopsy analysis was neither possible nor realistic. Further limitations of the study include: no control condition of 100 º knee flexion posture at 0 Hz was adopted; knee angles were visually inspected; finally no performance measures pre or post WBV were utilised. The reason for all of these potential limitations was due to time restraints in data collection of a large rugby squad completing a busy training camp. More importantly, the nature of the applied sport science consultancy work which formed this chapter included a remit given to determine which WBV frequency would elicit the highest EMG response.

An additional data analysis limitation was the use of only a band pass filter for EMG data. As shown by typical data in Figure 4.3 there appears evidence of motion artifacts related specifically to the fundamental frequency of WBV during an EMG recording. This is in agreement with previous work (Abercromby *et al.*, 2007a) who identified similar spikes in spectral analysis corresponding to the frequency of WBV exposure; and attributed them to electrical interference. A growing body of literature (Cormie *et al.*, 2006; Abercromby *et al.*, 2007a; Fratini *et al.*, 2009a; Fratini *et al.*, 2009b; Marin *et al.*, 2011) have utilised notch filtering techniques in an attempt to remove these artifacts. It is acknowledged that the data in this present study did not utilise these notch type filter technique and further research is warranted as the influence of such notch filters across different WBV frequencies is unknown. This will be discussed in further detail in chapter 6. Previous work has suggested that the magnitude of unfiltered EMG_{rms} data increase relative to WBV frequency (Fratini *et al.*, 2009b). It would be reasonable to hypothesis that 50 Hz would therefore produce the largest artifacts and therefore the highest EMG_{rms} amplitude values. This was not the case, suggesting further research is required regarding motion artifacts and filtering techniques in EMG signal recorded during WBV.

4.4.1 Conclusions

In conclusion no one single frequency of WBV elicited a highest response in EMG activity over another frequency. There appears some evidence to suggest that, in elite athletes, there may be an individualised EMG response to WBV frequency. This may be explained by inter participant variability in neuromuscular properties such as muscle fibre characteristics. However, further research is required to investigate this and to start to quantify the effect of a possible individualised response to WBV on performance.

Chapter 5 The role of frequency and expectancy effect on acute and individualised whole body vibration induced changes in electromyography and jump performance

5.1 INTRODUCTION

As shown in Chapter 4 there was no significant difference in EMG response during different WBV frequencies (no main effect for frequency reported in elite rugby athletes). Chapter 4 also reported large intra-participant variability (mean % CV of 47.5 ± 36.7 mV) in response to WBV frequencies (30, 35, 40 and 50 Hz); supporting previous literature (Di Giminiani *et al.*, 2009). This could be indicative of individualised responses to WBV frequency. As previously mentioned in section 4.1 an individualised jump performance response to chronic WBV frequency has been reported (Di Giminiani et al., 2009). However, the vast majority of WBV literature has utilised pre-determined frequencies, applied across all participants; therefore individuals may not have been exposed to their most effective WBV frequency. It is perhaps not surprising that the literature has reported conflicting findings for WBV responses on jump performance.

Previous literature (Delecluse *et al.*, 2005; Blottner *et al.*, 2006; Savelberg *et al.*, 2007; Belavy *et al.*, 2008; Moezy *et al.*, 2008) which manipulated WBV frequencies has often involved utilising different frequencies as progression in a chronic WBV programme setting (4 – 8 weeks in the literature examples given). These frequencies were pre-determined across all participants Other research has focussed on the application of different WBV frequencies and the effect on neuromuscular performance, or the effect on measured WBV parameters, for example head acceleration (Cardinale & Lim, 2003a; Fattorini *et al.*, 2006; Savelberg *et al.*, 2007; Kiiski *et al.*, 2008; Adams *et al.*, 2009). However, these pre-determined WBV frequencies were not individualised to each participant.

Di Giminiani *et al.* (2009) utilised EMG to record potential individualised responses to WBV frequency. Vastus lateralis EMG activity was recorded during isometric squat at varying WBV frequencies: 0; 20; 25; 30; 35; 40; 45; 50; and 55 Hz were obtained. A second anecdotal finding of the work by Di Giminiani *et al.* (2009) was intra-participant variability in EMG activity during the WBV frequencies. For example, one participant exposed to 45 Hz WBV, the EMG response was 74 % higher than that of 30 Hz exposure. Whereas, another participant exposed to 45 Hz WBV, the EMG response was 34 % less than that of 30 Hz (Di Giminiani et al., 2009). The frequency of WBV which elicited the highest EMG response was assigned as the participant's iWBV and subsequently used during a chronic WBV programme (8 weeks) for those participants in the individualised frequency group.

Compared to participants in a fixed frequency (30 Hz) WBV group, the iWBV group had a significantly greater improvement in squat jump height (3.1 and 0.7 cm improvements, representing 11 and 3 % improvements for iWBV and fixed frequency respectively). Continuous rebound jump power significantly improved (18 %) post programme only in the iWBV group; with no changes in the fixed frequency group (Di Giminiani et al., 2009). It appears that by individualising WBV frequencies, performance improvements following chronic WBV programme were greater. The use of continuous rebound jump performance can be argued a good indicator of explosive strength with similar reactive strength indices as other forms of jumps, such as drop jumps (Di Giminiani et al., 2009).

However, as previously mentioned in section 4.1, the study design was chronic in nature and participants were physically active, but not resistance trained. There appears a specific WBV response that may be related to training status in participants; speculated to involve muscle-tendon complex characteristics potentially influencing the amount of WBV transmission in those that are well-trained (Bullock et al., 2009). There are additional limitations worth outlining, such as a lack of drop jump performance as a measure of reactive strength and therefore, the influence of WBV on reactive strength index (RSI) is unknown.

It is also unclear whether Di Giminiani *et al.* (2009) utilised a volume-matched control blinded study design. It seems unlikely as only those in the iWBV group completed the individualised programme. Placebo and control aspects of study design highlight a general limitation with WBV literature. The role of placebo in WBV literature has not been addressed, as some research lacks any control group or control intervention (Adams et al., 2009). Those studies which did utilise a control group/intervention were often not volume-matched (Bullock et al., 2008). Even within volume-matched control studies, the placebo effect may not have been fully addressed; as during these control interventions the platform was not switched on. It is likely participants would have noticed this and a phenomenon associated with the placebo effect, the role of expectancy effect, may have an influence (McClung & Collins, 2007).

RSI has been developed as a measure of the explosive capabilities of the musculotendinous complex (Flanagan & Harrison, 2007; Flanagan *et al.*, 2008). The RSI is defined as jump height divided by ground contact time during a drop or depth jump (Young, 1995). The RSI measures the ability to quickly change from eccentric to concentric contractions and is seen as a reflection of neuromuscular characteristics of a fast stretch-shortening cycle (SSC) (Flanagan et al., 2008). To date there has only be very brief discussion regarding RSI and WBV (Di Giminiani *et al.*, 2009; Di Giminiani *et al.*, 2010); and no specific research into the response of RSI from varying WBV frequencies. As previously mentioned in section 2.3.6 muscle tuning may explain one of the mechanisms for WBV responses. This may vary due to individual's capacity to dampen the WBV stimulus (Cardinale & Lim, 2003b). Differences in muscle spindle properties, quantity and location of mechanoreceptors and visoelastic properties may contribute towards individuals response to WBV (Cardinale & Lim, 2003a). These differences are also likely to influence both jump height and ground contact time via neuromuscular strength and power generating capabilities; therefore influencing RSI.

The role of expectancy is represented by an individual's belief regarding the efficacy of an intervention (McClung & Collins, 2007). It has been found that, when the psychological effect during an intervention was isolated, a significantly improved performance was recorded (1.9 % decrease in 1000m running time trial) versus the purely interventional effect (McClung & Collins, 2007). Therefore, volume-matched control study designs could be influenced by the role of expectancy; as participants could perceive a more beneficial effect during WBV exposure than when they received no WBV. Those studies which tried to address this influence were Luo *et al.* (2008) and Luo *et al.* (2009). Both study designs involved a "sham" vibration protocol (0.02 µm D_{PTP} and 0.02 $m.s^{-2}$ acceleration). However, participants were informed that the control intervention was a sham vibration. Consequently participants may have viewed the sham exposures less beneficial as the "true" vibration exposure.

Previous literature (Adams *et al.*, 2009) has reported acute improvements in CMJ power following WBV durations of 30 – 60 s. This was using common WBV frequencies (30, 35, 40 and 50 Hz) and common D_{PTP} values (2 – 4 mm); however participants recruited were untrained and no neuromuscular analysis was completed on the pre-determined WBV frequencies. Further research is needed to investigate individualised neuromuscular responses to acute WBV frequency in well-trained participants, as to date the vast majority of existing literature has been in the chronic setting. This research should include explosive measures of neuromuscular performance aiming to identify potential WBV mechanisms. Such research also needs to be well controlled and address potential role of expectancy influences. Therefore, the objectives of this study were three fold:

1. To investigate the acute effects of WBV frequencies on the neuromuscular response of: EMG activity during WBV; and explosive jump performance pre-post WBV.
2. To determine whether an individualised response to frequency following acute WBV exists.
3. To characterise the role of expectancy effect during active WBV.

5.2 METHODS

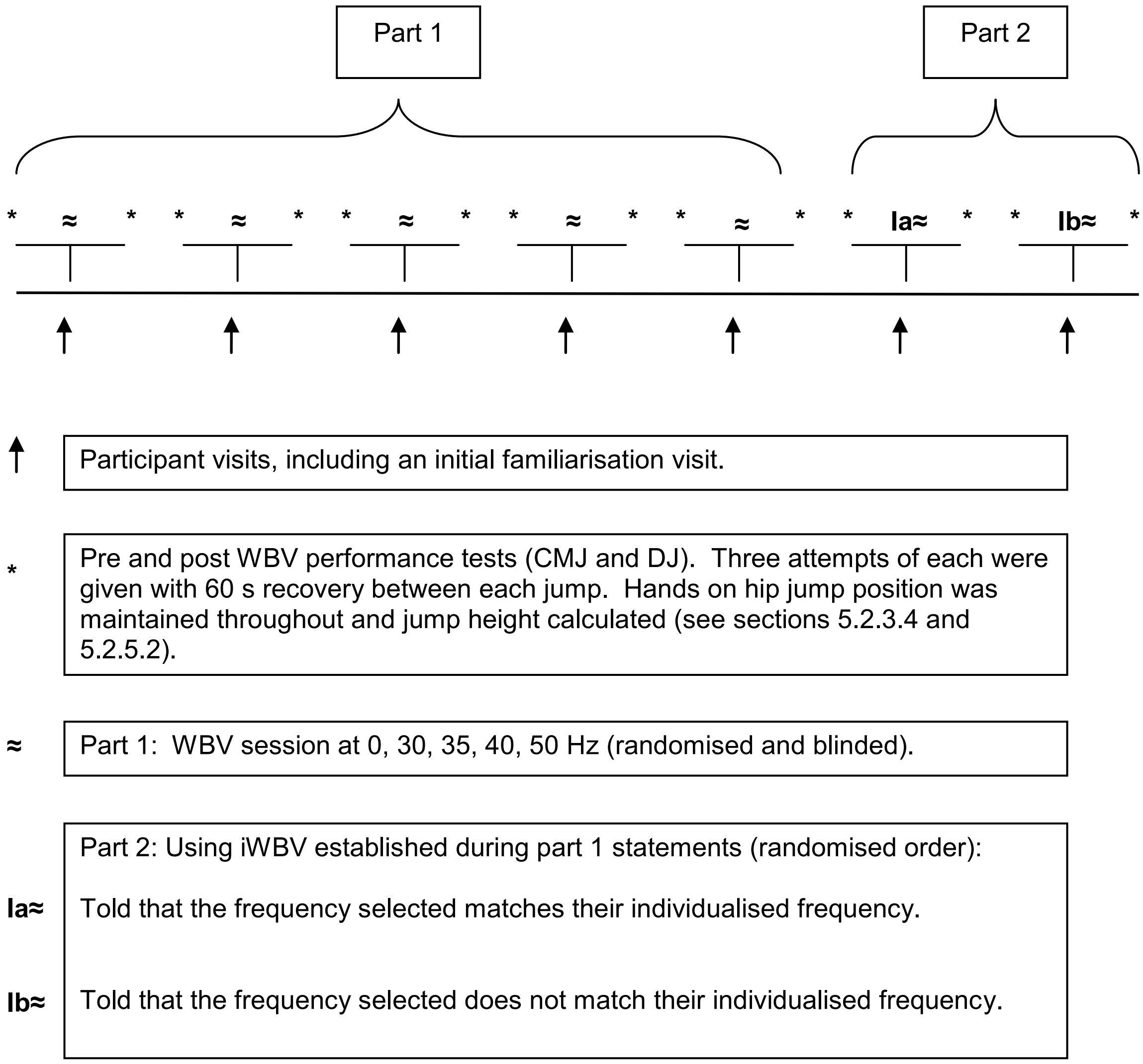
5.2.1 Experimental approach

The present study was divided into two parts. Part 1 focused on the effect of frequency on jump performance and EMG response, then subsequently determining whether individualised responses existed. Part 1 utilised a single blinded randomised repeated-measures design in which frequency was an independent variable and jump performance measures and EMG response were the dependent variables. Participants acted as their own control. Part 2 utilised a similar study design however, the independent variable was role of expectancy condition and dependent variables remained the same (i.e. frequency was fixed).

For part 1 participants attended five sessions in a randomised order, separated by at least 48 hours to allow full recovery and flexibility in arranging test sessions with participants' busy schedule. Each session participants experienced a different frequency of WBV (30, 35, 40 and 50 Hz with 0 Hz acting as a control condition). Before each WBV session participants were weighed (via force plate measurements) and completed a standardised body weight warm-up (see section 5.2.3.1). Participants then completed a standardised knee extension and knee flexion IMVC during which EMG data was recorded. For each session participants also completed pre WBV jump tests (CMJ and DJ) (see section 5.2.3.4). Participants were then exposed to 60 s of WBV at a randomly allocated frequency (adopting 90 ° knee flexion), during which EMG was recorded, before then repeating the same jump tests post WBV (see Figure 5.1). For each jump test, three attempts were completed and the best performance taken.

For part 2 participants were informed of their iWBV (the frequency which elicited the greatest positive response of CMJ height during part 1). Each participant was invited for a final two sessions. Both sessions involved identical pre and post WBV protocols as part 1. One of the two sessions each participant was informed they were receiving their iWBV, and actually received it. The other session, to explore the role of expectancy, each participant was informed they were receiving another WBV frequency other than their iWBV, to act as a control. Each participant still received their iWBV (see Figure 5.1).

Figure 5.1: Study design, part 1 and 2.



5.2.2 Participants

A total of 7 participants were recruited from this study (age 23 ± 4 years; height, 180 ± 6 cm; mass, 82.6 ± 9.0 kg). These participants fulfilled the inclusion criteria outlined in section 3.1. Participants were not informed of the second part of the study until completing the first part and were given the opportunity to take part if each participant wished to do so, following the same informed consent procedure.

5.2.3 Procedures

5.2.3.1 Warm up

A standardised warm-up was utilised. This was selected for obvious safety reasons as participants were asked to perform maximal efforts. The protocol was selected with both jump performance and EMG measures in mind. Protocols involving static stretching have been suggested to decrease (by 9.4 %) vertical jump performance compared to dynamic stretching; whereas the latter increased EMG activity (by 85 %) (Hough et al., 2009). Even though there is contradicting evidence that static stretching reduces EMG activity (Wallmann *et al.*, 2005; Wallmann *et al.*, 2008), this was for gastrocnemius EMG activity and not quadriceps and hamstrings, as measured in this present study. The use of dynamic stretching to conclude a warm up is recommended (Simic et al., 2012) suggesting static stretching can negatively influence maximal strength and explosive performance. Given the nature of the testing protocol, the warm-up protocol was modified from work investigating explosive jump performance reliability (Flanagan *et al.*, 2008; Ebben & Petushek, 2010), see below:

Five minutes cycle at self-selected speed (0.5 kg cycle load applied)
Slow BM squat x 5
Fast BM squat x 5
10 m forward lunges
10 m backward lunges
10 m lateral lunges
10 m walking quadriceps dynamic stretching
10 m walking hamstring dynamic stretching
CMJ increasing intensity up to 75 % over 5 reps

DJ increasing intensity up to 75 % over 5 reps

(Both jump components also acted as familiarisation for subsequent jump tests)

5.2.3.2 Electromyography

In addition to the general method section referring to EMG, the following procedure was completed per session across this study parts 1 and 2. EMG recordings were taken from right vastus lateralis, rectus femoris and biceps femoris as performance measures were all bilateral activities and due to the repeated-measures study design this was standardised. EMG data collection was completed during IMVC's and WBV exposure. During WBV, EMG was recorded between 20 to 30 s and 40 to 50 s time periods during a total of 60 s WBV exposure, for similar reasons as explained in section 4.2.3.

5.2.3.3 Isometric maximal voluntary contraction

Each session IMVC was completed pre jump tests and was standardised across parts 1 and 2 using the same strain gauge. Knee extension IMVC was obtained sitting on a weight bench at a 90 º knee angle (measured via visual goniometry) to simulate knee joint position subsequently adopted on the WBV platform, and also to elicit maximal voluntary activation (Becker & Awiszus, 2001). Participants placed their feet behind a cushioned fixed lever, allowing isometric knee extension to be undertaken. Knee flexion IMVC was performed in the same posture following a similar protocol with one exception, that participants placed their feet in front of a cushioned fixed lever, allowing isometric knee flexion to be undertaken. During both IMVC tests the upper body position was standardised via hand placement under the bench and trunk movement was minimised, although not fixed via straps for example.

5.2.3.4 Jump performance tests

To investigate the effect of WBV on explosive jump performance two jump tests were selected. Potential mechanisms responsible for positive jump performance post WBV have been proposed which have included stretch reflex and/or SSC mechanisms (Cardinale & Bosco, 2003). Therefore, appropriate jump tests were required. CMJ was selected as a measure of slow SSC (Young, 1995; Linthorne, 2001) and as a measure of both neuromuscular contractile and elastic properties which WBV stimulus may influence (Adams et al., 2009). However, SSC may not contribute significantly to CMJ performance (Reiser et al., 2006). Therefore, a second jump test was selected, drop jump (DJ). This is characterised by a faster SSC (Young, 1995) and allows a measure of musculotendinous complex explosive capabilities to be quantified and expressed as RSI (Flanagan et al., 2008). In addition, DJ is the only jumping test which allows ground contact time to be identified, a component in determining RSI (Ebben & Petushek, 2010). Participants were instructed to wear suitable footwear throughout pre and post jump tests and to ensure the same footwear was worn for each test session.

Pre WBV CMJ tests were completed for each session participants attended. Instructions were given to maintain hands on hips throughout. This was to standardise arm movement during CMJ as this can significantly alter CMJ performance (Harman *et al.*, 1990; Lees *et al.*, 2004), as well as increase inaccuracies when computing jump height (Linthorne, 2001). Participants were informed to stand and, on command, dip to a self-selected level and complete a maximal CMJ landing on both feet. Participants completed three CMJ performances with 60 s recovery between each attempt (Read & Cisar, 2001). The best jump height of the three attempts was utilised in subsequent calculations. Post WBV CMJ tests were completed within 60 s of WBV exposure. This was to ensure any potential beneficial effects on jump performance were recorded, as positive effects on CMJ power following WBV may be short lived; peaking at 60 s (Da Silva-Grigoletto et al., 2011) and remaining significant 5 minutes post WBV, but not 10 minutes post WBV (Adams et al., 2009). Post WBV CMJ used an identical protocol as pre tests.

Participants also completed DJ tests pre and post WBV immediately after CMJ tests. Hands on hips standardisation was maintained as arm involvement, specifically during DJ, can influence jump height (Laffaye et al., 2006). Participants stood on a standardised 40 cm high box which is within a range of optimal heights for DJ performance (Read & Cisar, 2001; Laffaye & Choukou, 2010). Participants were instructed to: step off, not jump; land with both feet; jump as fast and as high as possible; and land on the same location. Participants were given 60 s recovery between attempts as this has been suggested as sufficient rest between one DJ maximal effort (Read & Cisar, 2001). Each participant completed three pre and three post WBV attempts. Best jump height and quickest contact time were subsequently used in calculations.

5.2.3.5 Whole body vibration

In addition to the general methods section detailing the WBV platform (VM Master) and WBV parameters, this section will summarise the protocol for parts 1 and 2. For part 1 participants were exposed to five different WBV frequencies (0, 30, 35, 40 and 50 Hz) for 60 s duration and 3 mm D_{PTP}.

The frequency range was chosen based on previous literature which found: 30 Hz to elicit the highest vastus lateralis EMG activity (Cardinale & Lim, 2003b); 40 Hz to elicit the highest jump performance improvements (Turner et al., 2011). In addition based on previous work (Chapter 4) both 35 and 50 Hz were among WBV frequencies which individuals elicited the greatest EMG response. Finally, 30 to 50 Hz represented a common range of frequencies utilised across WBV platform manufacturers. The sequence of frequencies was randomised and single blinded to counteract any order effect and to allow subsequent investigation into the role of expectancy in part 2. Participants stood barefoot at 90 º knee flexion throughout the 60 s of WBV, monitored visually. This is to ensure the quadriceps were placed under a degree of stretch, (for further details see general methods section 3.3).

For part 2 the WBV was set at the individualised frequency based on results from part 1. The participant was told certain information regarding frequency, which will be detailed in the following section; otherwise, the protocol was identical to part 1.

5.2.3.6 Role of expectancy

Of the 7 participants who completed part 1, 6 participants (age, 23 ± 4 years; height, 180 ± 6 cm; weight, 86.8 ± 9.7 kg) volunteered to undertake part 2. The one participant who declined did so due to university time pressures and not due to any adverse effects of completing part 1. As shown in Figure 5.1 participants attended two sessions. For the first session participants were told they would be receiving their IWBV and actually received it, using the following script:

"*This session of WBV will use your individualised frequency. As you may remember this was the frequency which gave you the best jump performance out of all of the 5 frequencies used. This means that this session of WBV will improve your jump performance. You should give maximum effort during the jumping tests."*

Based on McClung & Collins (2007).

For the second session participants were told they would receive a frequency different to that of their IWBV as a control. However, participants still received their IWBV frequency, using the following script:

"As you know the aim of this study is to investigate the effect of giving an individualised frequency of WBV on jump performance. So to compare we need to give you a frequency of WBV which is not your individualised frequency. This visit will use one of the other frequencies you experienced during the first 5 visits. This means that this session of WBV may not improve your jump performance as much as

the other visit which used your individualised frequency. But you should still give maximum effort during the jumping tests."

Based on McClung & Collins (2007).

Immediately after completing the full protocol for part 2, participants were fully debriefed on the true protocol and the reasons why. Participants were offered opportunity to ask any questions. None of the participants appeared concerned at the apparent "misleading" protocol and were happy after the explanation was given.

5.2.4 Equipment

5.2.4.1 Electromyography

For detailed description of the EMG equipment utilised for chapter 5, please see section 3.2.2.

5.2.4.2 Isometric maximal voluntary contraction

Participants completed both IMVC knee extension and IMVC knee flexion sitting on a knee extension weights bench (Powersport, Bridgend, UK). For IMVC knee extension metal chains were utilised to fixed the cushioned level allowing isometric knee extension to be performed (Figure 5.2). Between these chains a strain gauge was fitted (Globus strain gauge, Globus Italia, Italy). The strain gauge was attached to a processing unit (Globus ergometer, Globus Italia) which calculated maximum force generated during IMVC knee extension (kg). The strain gauge was calibrated against a standardised weight (10 kg). For IMVC knee flexion participants performed knee flexion against the fixed lever.

Figure 5.2: Bench equipment set up for isometric maximal voluntary contraction for both knee extension (using the illustrated strain gauge) and knee flexion.

5.2.4.3 Jump performance tests

Both jumps (CMJ and DJ) utilised a force plate (Kistler®, Model 9281B12, Kistler Instrument Corp, USA). Data from the force plate was sent via an amplifier (Kistler® Model 9865B 8 channel charge amplifier, Kistler Instrument Corp, USA) to a PC where data was collected and processed using software (Bioware® for windows version 3.24, Kistler Instrument Corp, USA). Participants' BM was recorded before jump tests were performed. This had two purposes, firstly to obtain participants' BM each session and secondly to calibrate the force plate to each participant for subsequent kinematic data analysis (see section 5.2.5.2).

Use of force plates to calculate ground reaction forces and jump performance has been referred to as a criterion method (Buckthorpe et al., 2012) and has been recommended for accurate evaluation of jump performance (Dugan et al., 2004). For

DJ performance tests a plastic step up was utilised at a height of 40 cm. This was within optimal box height for DJ performance (Read & Cisar, 2001; Laffaye & Choukou, 2010). This was placed in a standard position in relation to the force plate allowing participants to step off and comfortably land on the force plate.

5.2.4.4 Whole body vibration

As the WBV was delivered by the same WBV platform (VM Master, VibraMachines Ltd, Ledborough, UK) across multiple studies, details of parameters are discussed in section 3.3.

5.2.5 Data analysis

5.2.5.1 Electromyography

As the majority of EMG data analysis is similar across studies description can be found in section 3.2.3. However, there are aspects of EMG data analysis specific to this study, which will be outlined below.

EMG data collected during WBV was electrically filtered via a band pass filter (20 – 450 Hz) as explained in section 3.2.3 (Luo et al., 2008). As already mentioned EMG data collection during WBV was completed in two sets of 10 s samples (20 – 30 s and 40 – 50 s). Before subsequent EMG data analysis both samples were compared. Mean values for both samples were calculated and compared via paired *t*-tests across all three muscles. No significant differences were found between sets of 10s samples of EMG_{rms} for vastus lateralis (p = 0.15), rectus femoris (p = 0.38) or bicep femoris (p = 0.89). Subsequently, the mean from both 10 s samples was calculated and used for further analysis.

As this study was a repeated-measures design EMG data was collected over several different sessions. Inevitably there will be variability in EMG magnitude and a method to reduce this across different test sessions is via a normalisation process

(Enoka, 2002); often against a standard magnitude of EMG. Therefore, EMG data collected during IMVC knee extension and IMVC knee flexion formed the standard response all other EMG data was normalised to (nEMG). Vastus lateralis and rectus femoris EMG data was recorded during WBV, normalised against IMVC knee extension EMG data; and bicep femoris EMG data recorded during WBV was normalised against IMVC knee flexion EMG data.

5.2.5.2 Jump performance tests

Jump height of both CMJ and DJ was calculated as an indication of stiffness and performance (Butler et al., 2003).

For CMJ height, flight time (f_{time}) was identified from vertical ground reaction force (VGRF) curves (e.g. Figure 5.3). Initiation and completion of f_{time} (VGRF < 10 N and > 10 N respectively) was identified (f and g, Figure 5.3). The time period between these two points was designated as f_{time} (h, Figure 5.3). CMJ height was calculated using the following equation:

$$\text{CMJ height (m)} = \frac{g \times f_{time}^{2}}{8}$$

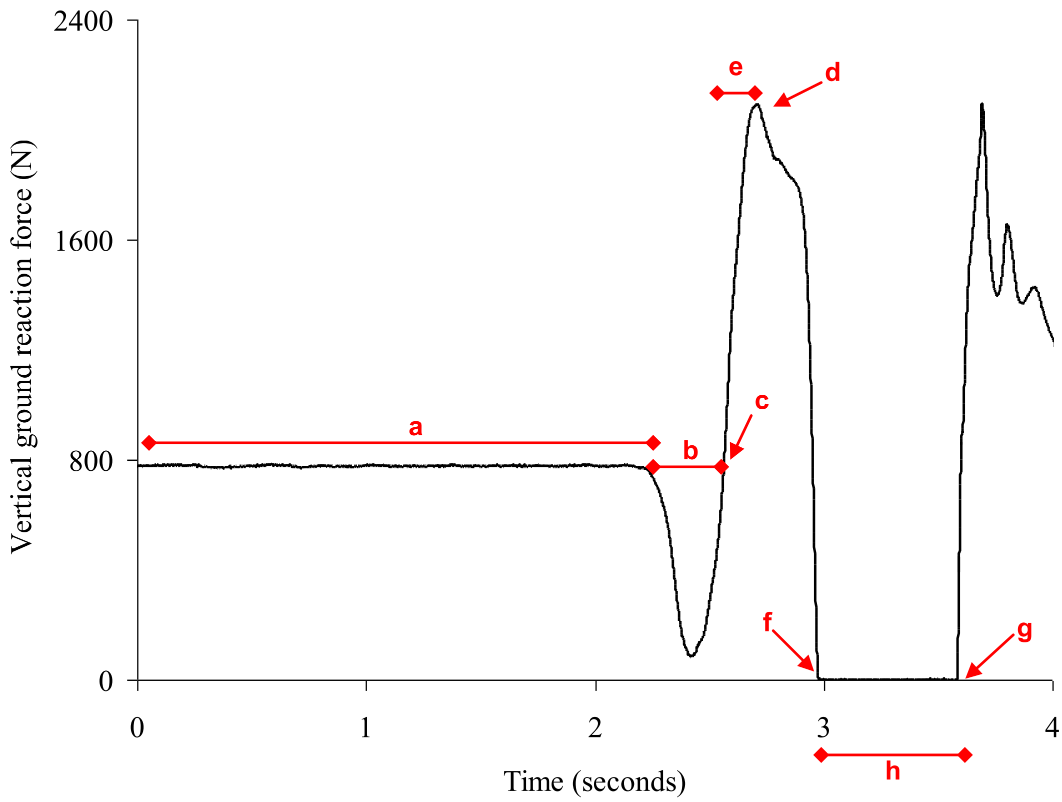
g = gravity (9.81 ms^{-2})

f_{time} = flight time

CMJ height in metres was then converted to centimetres

Adapted from Bosco *et al.* (1998)

Figure 5.3: Example of vertical ground reaction force curve of countermovement jump. Label a, participant stationary on force plate, BM determined; label b, unweighing phase; label c, VGRF, equals BM, identifying end of unweighing phase; label d, peak force during CMJ concentric phase; label e, RFD defined as time period from end of unweighing phase to peak force; label f, take off point VGRF below 10 N threshold; label g, landing point VGRF above 10 N threshold; label h, f_{time} defined as time period between take off and landing (Bosco & Komi, 1979; McLellan *et al.*, 2011).

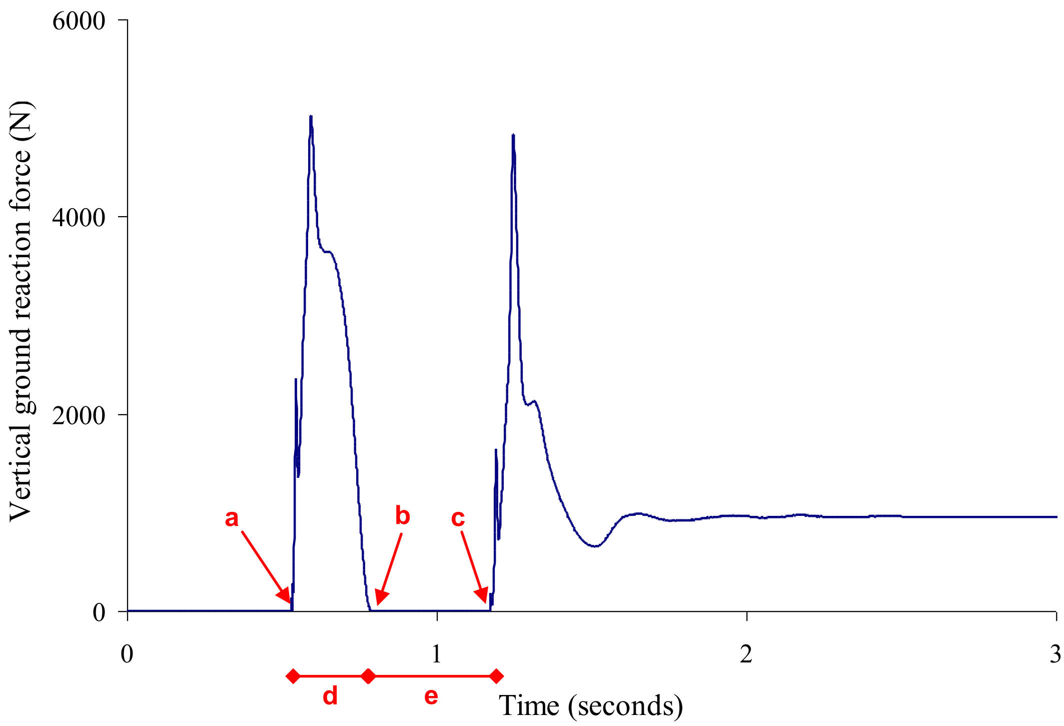


CMJ peak force was also identified from VGRF curves as the maximum amount of force generated before take off (Bosco & Komi, 1979). To calculate RFD first the unweighing phase of CMJ was identified (Bosco & Komi, 1979) (b, Figure 5.3). RFD was calculated as the peak force over the time to generate peak force (McLellan et al., 2011) (e, Figure 5.3).

The rationale for selecting these parameters was due to CMJ variables such as CMJ height and peak force representing independent characteristics of lower limb muscular CMJ performance (Young et al., 2011).

Regarding DJ data a similar approach using VGRF curves was utilised. Initial foot contact, take off and second foot contact events were identified using the same 10 N threshold (see Figure 5.4). DJ height in metres was calculated using f_{time} as already explained (Bosco et al., 1998). Ground contact time (CT) was determined as the time period between initial foot contact and take off events (d, Figure 5.4). RSI was calculated as DJ height (in metres) divided by CT (in seconds) and is a reliable measure of musculotendinous explosive capabilities (Flanagan et al., 2008).

Figure 5.4: Example of vertical ground reaction force curve of countermovement jump. Label a, initial foot contact; label b, take off point; label c, second foot contact landing; label d, ground contact time; label e, f_{time}.



5.2.5.3 Change in jump performance measures

Pre – post change in jump performance measures were calculated for 0 Hz (Δ0 Hz) which acted as a control condition to compare against participants best change pre to post WBV (Δbest across 30, 35, 40 or 50 Hz).

5.2.5.4 Electromyography and countermovement jump height change

The largest positive difference in vastus lateralis and rectus femoris EMG_{rms} versus 0Hz was calculated across 30, 35, 40 and 50 Hz. This value was used for analysis of correlation against the Δbest in CMJ height.

5.2.6 Statistical analysis

Vastus lateralis and rectus femoris peak EMG_{rms} during IMVC knee extension; biceps femoris peak EMG_{rms} during IMVC knee flexion; peak force during IMVC; and finally jump performance measures pre WBV (CMJ height, CMJ peak force, RFD during CMJ and RSI during DJ) were subjected to measures of reliability (Table 5.1). These included determining mean and standard deviation(SD), the change in mean, confidence levels (%), intraclass correlation coefficient (ICC) and coefficient of variation (% CV) using methods advocated by Hopkins *et al.* (2000).

Table 5.1: Reliability data for EMG_{rms} during isometric maximal voluntary contraction (IMVC) and reliability data for performance variables. Vastus lateralis and rectus femoris peak EMG_{rms} during IMVC knee extension; biceps femoris peak EMG_{rms} during IMVC knee flexion. IMVC, peak force during IMVC; CMJ, countermovement jump; peak force, peak force during countermovement jump; RFD, rate of force development; RSI, reactive strength index; mV, millivolts; kg, kilograms; cm, centimetres; N, newtons; $N.s^{-2}$, Newton per second; ± SD, standard deviation; confidence levels (%); ICC, intraclass correlation coefficient; % CV, coefficient of variation.

Variable	**Mean (± SD)**	**Change in mean**	**Confidence limits**	**ICC**	**% CV**
Vastus lateralis EMG (mV)	0.330 ± 0.180	-0.011	-0.135 to 0.113	0.74	28.3
Rectus femoris EMG (mV)	0.235 ± 0.263	0.041	-0.133 to 0.215	0.44	41.6
Biceps femoris EMG (mV)	0.168 ± 0.138	-0.003	-0.123 to 0.113	0.50	44.3
IMVC (kg)	102.2 ± 23.0	-4.763	-77.13 to 39.04	0.77	11.0
CMJ height (cm)	36.6 ± 6.5	-2.8	-2.4 to 1.0	0.94	4.8
Peak force CMJ (N)	2114.6 ± 408.1	10.7	-103.9 to 125.4	0.96	4.0
RFD CMJ ($N.s^{-2}$)	6617.3 ± 5011.5	114.8	-2427.1 to 2656.0	0.86	32.4
DJ RSI	0.96 ± 0.20	0.0	-0.16 to 0.16	0.65	12.7

The effect of different WBV frequencies on EMG_{rms} response of three selected muscles were analysed by means of a 1-way repeated-measures ANOVA (frequency [0, 30, 35, 40, 50 Hz]). The same statistical approach was completed on nEMG data when analysing the effect of different WBV frequencies.

The effect of different WBV frequencies on jump performance measures was analysed by means of a 2-way repeated-measures ANOVA (frequency [0, 30, 35, 40, 50 Hz] x time [pre and post]). If a significant interaction effect was found a 1-way ANOVA was utilised to further explore interactions.

Possible individual responses to frequency (pre-post WBV change scores Δ 0 Hz versus Δ best) were analysed by means of a paired *t*-test with Bonferroni correction. Analysis of correlation between largest positive difference in vastus lateralis and rectus femoris EMG_{rms} and CMJ height response of corresponding WBV frequency were completed via 2-tailed Pearson Correlations. The influence of role of expectancy on CMJ height was analysed via a 1-way repeated-measures ANOVA [IWBV, known, misled].

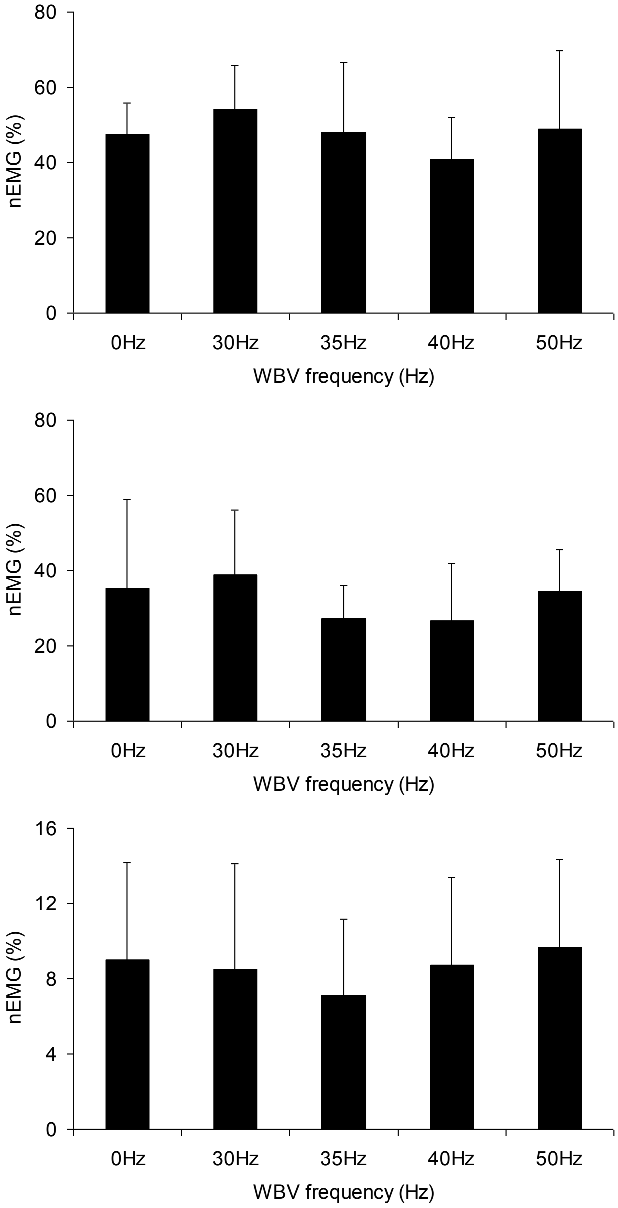
5.3 RESULTS

5.3.1 Electromyography data during whole body vibration

5.3.1.1 Normalised electromyography

No significant main effect for frequency was found for: vastus lateralis $nEMG_{rms}$ (F [4,24] = 0.86, p = 0.50) with a moderate effect size (r = 0.36); rectus femoris $nEMG_{rms}$ (F [4,24] = 1.20, p = 0.34) with moderate effect size (r = 0.41); and for bicep femoris $nEMG_{rms}$ (F [4,20] = 0.44, p = 0.78) with small effect size (r = 0.28), Figure 5.5.

Figure 5.5: Mean ± SD $nEMG_{rms}$ of three muscles, during WBV at different frequencies, no significant main effect for frequency was found.



A: Vastus lateralis $nEMG_{rms}$, (normalised against IMVC knee extension).

B: Rectus femoris $nEMG_{rms}$, (normalised against IMVC knee extension).

C: Biceps femoris $nEMG_{rms}$, (normalised against IMVC knee flexion).

5.3.1.2 Mean electromyography

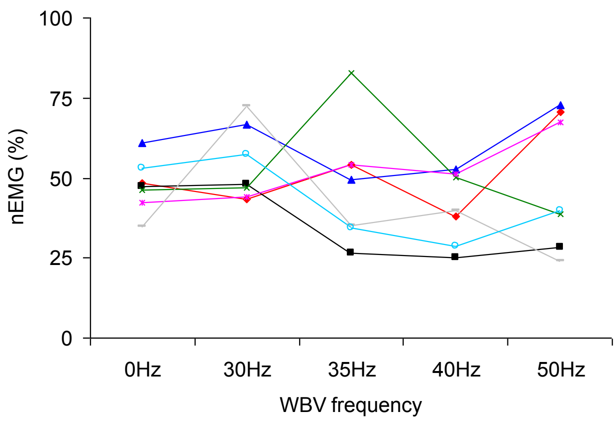
No significant main effect for frequency was found for: vastus lateralis EMG_{rms} (F [4,24] = 0.12, $p = 0.98$) with small effect size ($r = 0.14$); rectus femoris EMG_{rms} (F [4,24] = 0.26, $p = 0.90$) with small effect size ($r = 0.20$) and bicep femoris EMG_{rms} (F [4,24] = 0.44 $p = 0.78$) with small effect size ($r = 0.26$).

5.3.1.3 Individual responses in normalised electromyography

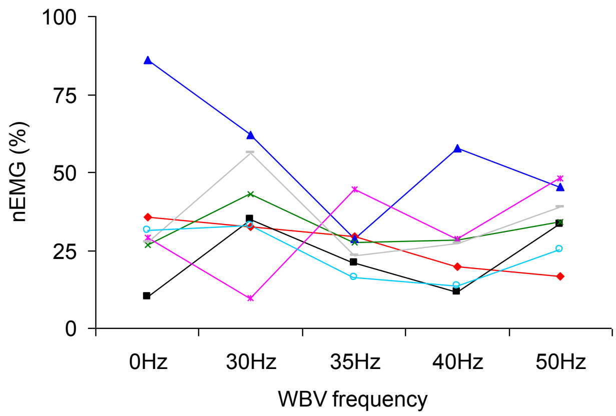
Individual responses during different frequencies of WBV are shown in Figure 5.6, which displays $nEMG_{rms}$ data for each participant across five WBV frequencies.

Figure 5.6: nEMG (%) during WBV at five different frequencies for all participants.

A: vastus lateralis

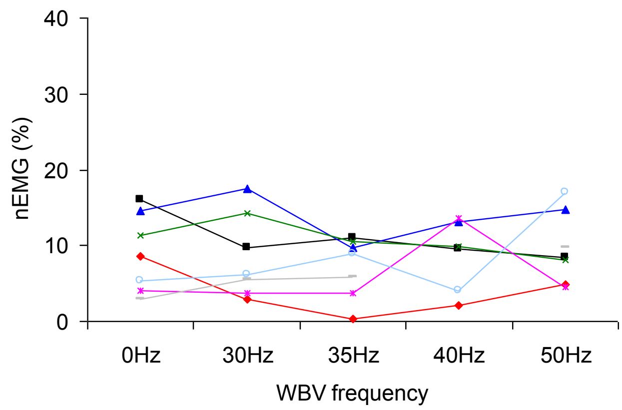


B: rectus femoris



C: biceps femoris

One participant's (grey) bicep femoris EMG data during 40 Hz WBV failed to record and thus is not presented in graph C.

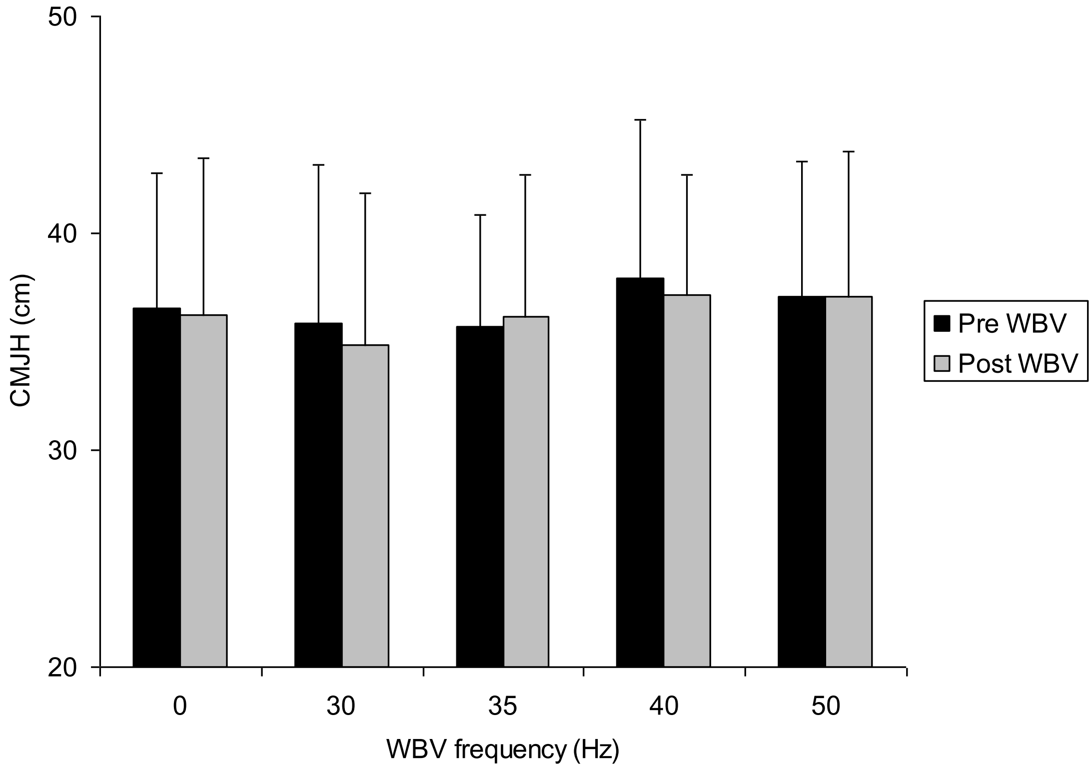


5.3.2 Performance measures pre and post whole body vibration

5.3.2.1 Countermovement jump height

No significant main effect for frequency was found (F [4,24] = 1.45, p = 0.25) with moderate effect size (r = 0.45). No significant main effect for time was found (F [1,6] = 1.11, p = 0.33) with moderate effect size (r = 0.40). No significant frequency x time interaction was found (F [4,24] = 0.55, p = 0.70) with small effect size (r = 0.28), Figure 5.7.

Figure 5.7: Mean ± SD CMJ height (cm) pre and post WBV at five different frequencies. No significant main effects for frequency or time, or significant frequency x time interaction effects were found.

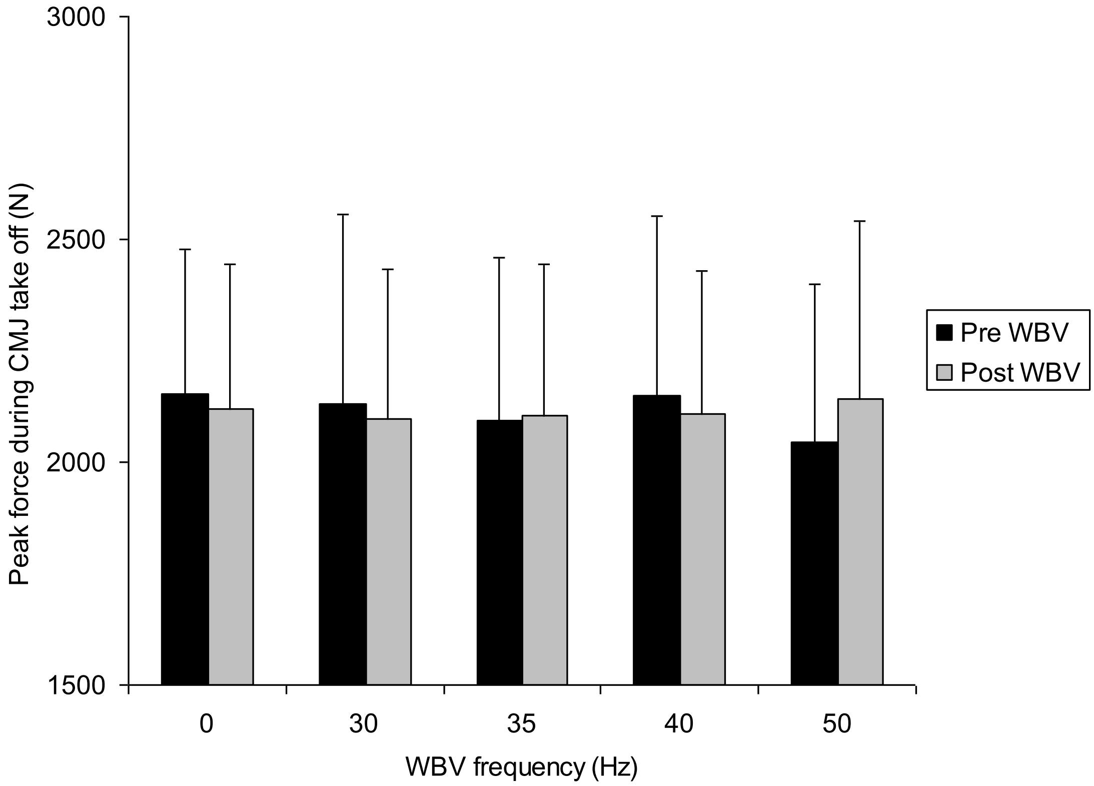


5.3.2.2 Peak force during countermovement jump take off phase

No significant main effect for frequency was found (F [4,24] = 0.37, p = 0.83) with small effect size (r = 0.24). No significant main effect for time was found

(F [1,6] = 0.002, p = 0.96) with small effect size (r = 0.14). No significant main effect for frequency x time interaction was found (F [4,24] = 2.31, p = 0.09) with large effect size (r = 0.53), Figure 5.8.

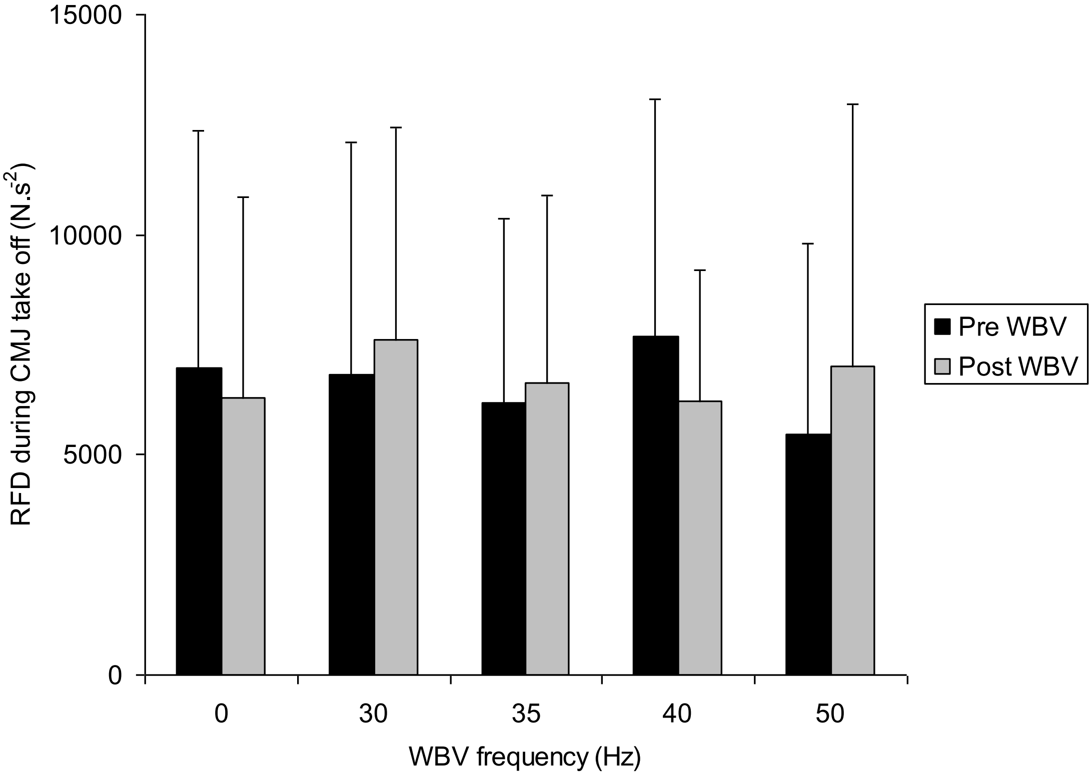
Figure 5.8: Mean ± SD peak force (N) during CMJ take off phase pre and post WBV at five different frequencies. No significant main effects for frequency or time, or significant frequency x time interaction effects were found.



5.3.2.3 Rate of force development during countermovement jump take off phase

No significant main effect for frequency was found (F [4,24] = 0.80, p = 0.54) with moderate effect size (r = 0.35). No significant main effect for time was found (F [1,6] = 0.21, p = 0.67) with small effect size (r = 0.17). No significant frequency x time interaction was found (F [4,24] = 1.86, p = 0.15) with moderate effect size (r = 0.49), Figure 5.9.

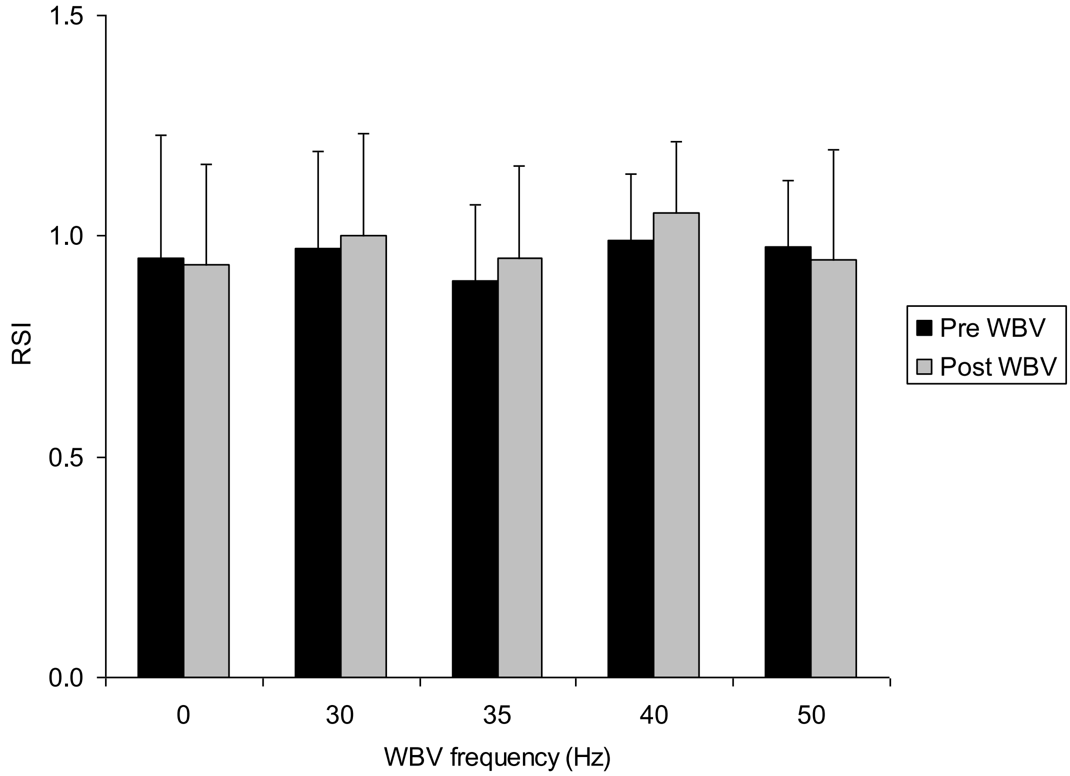
Figure 5.9: Mean ± SD RFD ($N.s^{-2}$) during CMJ take off phase pre and post WBV at five different frequencies. No significant main effects for frequency or time, or significant frequency x time interaction effects were found.



5.3.2.4 Reactive strength index

No significant main effect for frequency was found (F [4,24] = 0.64, p = 0.64) with moderate effect size (r = 0.32). No significant main effect for time was found (F [1,6] = 1.58, p = 0.26) with moderate effect size (r = 0.46). No significant frequency x time interaction was found (F [4,24] = 0.81, p = 0.53) with moderate effect size (r = 0.45), Figure 5.10.

Figure 5.10: Mean ± SD RSI pre and post WBV at five different frequencies. No significant main effects for frequency or time, or significant frequency x time interaction effects were found.



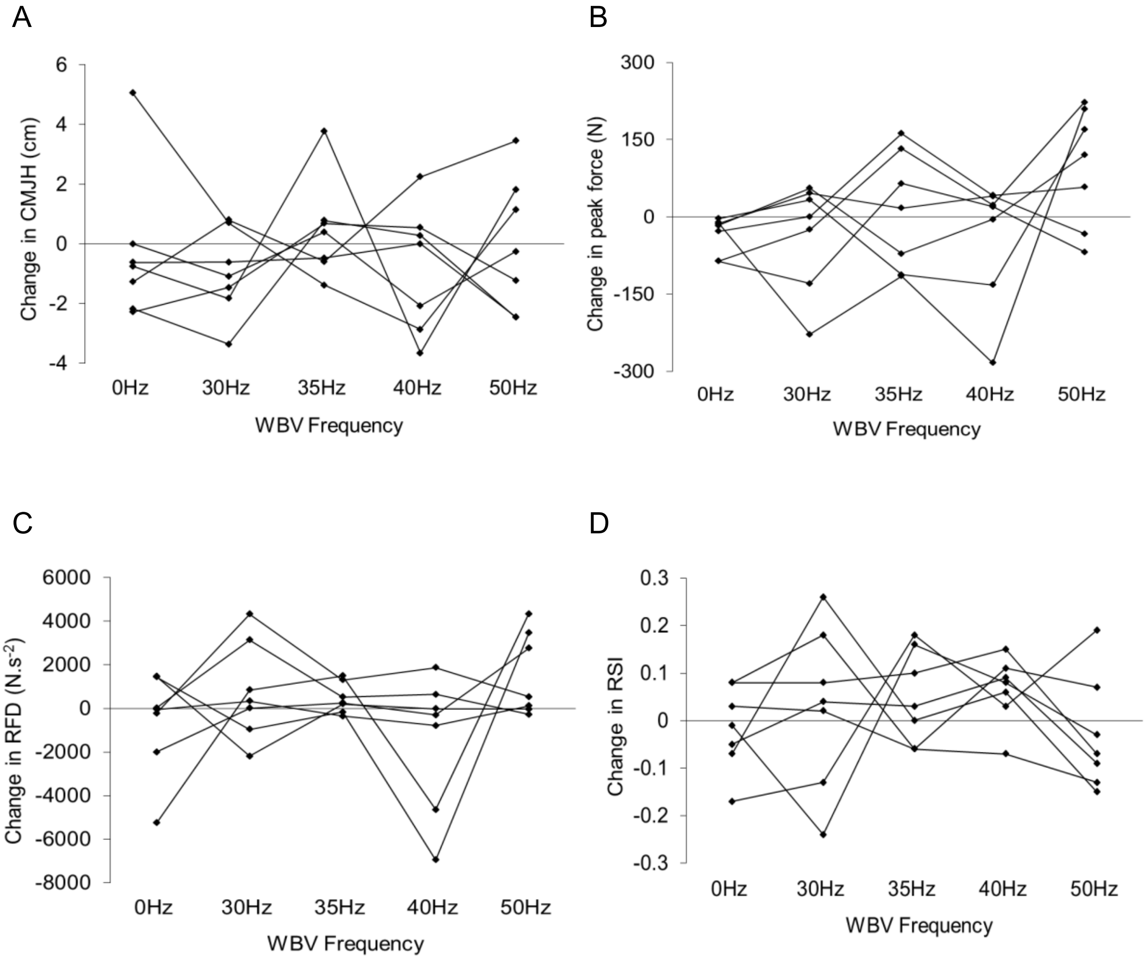
5.3.3 Individual responses in jump performance measures

No significant difference was found between change pre – post WBV at 0 Hz (Δ 0 Hz) and participants' best change pre – post WBV (Δ best) in CMJ height (p = 0.18). For peak force generated during CMJ Δ best was significantly greater than Δ 0 Hz (t [6] = -6.09, $p < 0.01$). For RFD, Δ best was significantly greater than Δ 0 Hz (t [6] = -2.49, $p < 0.05$). For RSI, Δ best was significantly greater than Δ 0 Hz (t [6] = -3.24, $p < 0.05$), Table 5.2. Individual change across participants is illustrated in Figure 5.11.

Table 5.2: Change pre – post WBV in performance measures for 0Hz (Δ 0Hz) against each participants' best change pre – post WBV (Δ best) and largest positive difference in vastus lateralis and rectus femoris EMG_{rms} versus 0Hz. Peak force, peak force during countermovement jump take off phase; RFD, rate of force development during countermovement jump take off phase; RSI, reactive strength index; VL, vastus lateralis; RF, rectus femoris; SD, standard deviation. * denotes significantly greater best change versus 0Hz change.

	CMJ height (cm)		Peak force (N)		RFD ($N.s^{-2}$)		RSI		EMG vs. 0Hz (mV)	
	Δ 0Hz	Δ best	Δ 0Hz	Δ best	Δ 0Hz	Δ best	Δ 0Hz	Δ best	VL	RF
P1	-0.63	0	-27.9	162.0	-215.2	4321.1	-0.05	0.09	0.03	0
P2	-2.29	0.68	-85.9	222.8	-2004.6	2766.2	-0.07	0.26	0.09	0.04
P3	-1.28	3.45	-11.0	209.3	1438.5	3457.8	0.08	0.18	0	-0.01
P4	5.05	1.14	-85.8	64.6	1467.9	202.7	-0.01	0.16	0.09	0.02
P5	-2.18	0.78	-13.4	46.1	23.3	3144.9	0.08	0.15	0.09	0
P6	-0.75	3.78	-3.6	169.7	-5260.1	4333.4	-0.17	0.19	0.10	0.05
P7	0.00	0.39	-16.9	120.2	-46.5	334.8	0.03	0.02	0.03	0
Mean	-0.30	1.46	-34.9	142.1*	-656.7	2651.6*	-0.02	0.15*	0.06	0.01
SD	2.50	1.52	35.5	68.2	2343.1	1726.6	0.09	0.08	0.04	0.02

Figure 5.11: Individual change in performance measures (pre-post WBV) at five different frequencies. Change pre – post WBV in performance measures: A, countermovement jump height (CMJH). B, peak force during countermovement jump take off phase. C, rate of force development (RFD) during countermovement jump take off phase. D, reactive strength index (RSI).



The frequencies of WBV which elicited the highest response in CMJ height and the largest positive difference in vastus lateralis and rectus femoris for each participants are presented in Table 5.3.

Table 5.3: Participants WBV frequencies eliciting the highest CMJ height response (pre-post) and largest positive difference in vastus lateralis and rectus femoris EMG_{rms}.

Participant	CMJ height response frequency	Vastus lateralis frequency	Rectus femoris frequency
1	40Hz	50Hz	35Hz
2	35Hz	30Hz	50Hz
3	50Hz	50Hz	30Hz
4	35Hz	50Hz	35Hz
5	35Hz	50Hz	40Hz
6	35Hz	30Hz	30Hz
7	50Hz	50Hz	40Hz

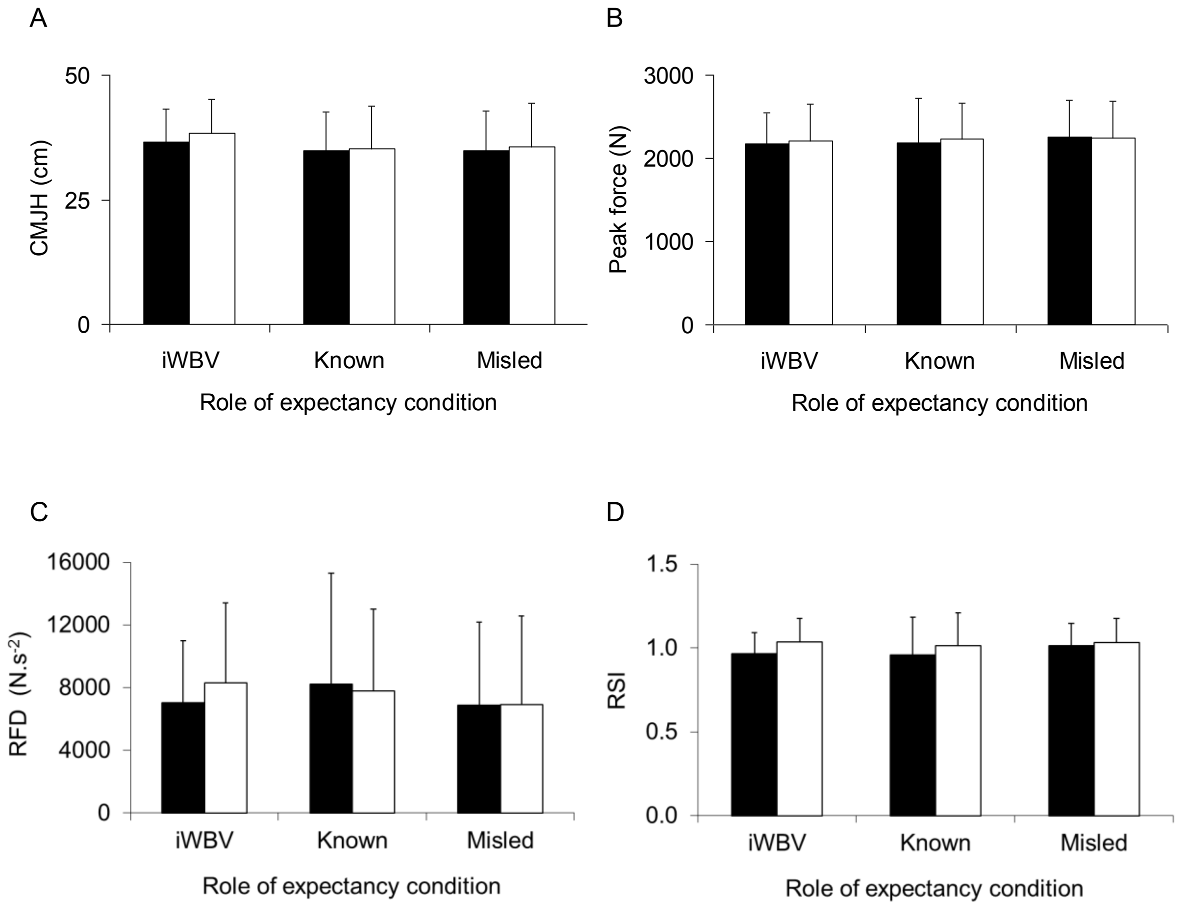
5.3.4 Electromyography and countermovement jump height correlation

There were no significant relationships between largest positive difference in EMG_{rms} versus 0 Hz and Δ best in CMJ height (pre to post WBV) (range of r = -0.43 – 0.17 and p = 0.34 – 0.72 for vastus lateralis and range of r = -0.74 – 0.25 and p = 0.06 – 0.83 for rectus femoris).

5.3.5 Role of expectancy effect in performance measures

No significant main effects for role of expectancy condition were found in CMJ height, peak force, RFD or RSI (p = 0.34 – 0.80) with small to moderate effect sizes (r = 0.20 – 0.44). No significant main effects for time were found in CMJ height, peak force, RFD or RSI (p = 0.08 – 0.58) with moderate to large effect sizes (r = 0.32 – 0.74). No significant role of expectancy condition x time interactions was found in the same performance measures (p = 0.29 – 0.67) with small to moderate effect sizes (r = 0.28 – 0.47) (Figure 5.12).

Figure 5.12: Role of expectancy effect in: A, CMJH; B, peak force; C, RFD; and D, RSI, comparing individualised frequency of WBV with known and misled role of expectancy conditions. iWBV, individualised frequency of WBV as identified by best response in performance following 30, 35, 40 and 50Hz; known, participants received iWBV and were informed; misled, participants received iWBV however were informed another frequency was utilised.



5.4 DISCUSSION

The objectives of this present study were three fold: to investigate the acute effects of WBV frequency on EMG and explosive jump performance; to determine whether individualised responses of EMG activity (during WBV) and jump performance (pre-post WBV) to frequency exists in trained participants; and finally, to characterise a potential role of expectancy effect during acute WBV. The main findings were firstly, frequency did not appear to consistently influence overall EMG activity or jump performance responses across participants as a group. Secondly, there is evidence to suggest an individualised response of EMG activity to WBV frequency. Compared to a control, the frequency eliciting the best change (pre-post WBV) in jump performance was significantly higher than the change (pre-post) elicited by 0 Hz. This was true for all jump performance outcome measures except for CMJ height. Finally, there appears to be little or no role of expectancy effect during acute WBV on any of the jump performance measures.

5.4.1 The effect of whole body vibration frequency on electromyography

The finding that WBV frequency did not influence overall EMG activity (either vastus lateralis, rectus femoris or bicep femoris) response across participants as a group supports previous literature (Hopkins *et al.*, 2009; Hannah *et al.*, 2012). The finding also supports the previous initial work (see result section 4.3) in respect that, overall no frequency main effect was reported. It should be highlighted at this point; that the overall statistical approach regarding the EMG response to varying WBV frequencies taken by this present study may under-represent an individualised response. It is acknowledged that, while no main effect for frequency was found across all participants, that at an inter-participant level, individualised responses to WBV frequency may exist and go undetected by this approach. These aspects will be discussed later in this section. It is noteworthy that the present findings support that of chapter 4, even with differences in WBV platform type and thus vibration characteristics (vertical versus tri-planar; and 3 versus 4 mm D_{PTP} respectively) and in trained status and strength of the participants..

This conflicts with Cardinale & Lim (2003b) as already discussed in section 4.4. However, when compared to this present study further possible reasons for the conflict emerge. Cardinale & Lim (2003b) utilised international female volleyball participants; whereas, this present study recruited university level male participants (70 % of whom were rugby players, 30 % of whom were volleyball players). Gender specific responses to WBV may account for the conflict due to the difference in female versus male participant recruitment (Bazett-Jones et al., 2008a). The training history of the athletic participants between studies may also influence the response to

WBV. However, the type of training even at university level that a rugby player would complete is likely to be very different to that of international volleyball players. Rugby players would emphasise predominately strength type training; whereas volleyball players are likely to emphasise more plyometric type training. As a result the different groups of participants are likely to have different neuromuscular recruitment strategies. For rugby players this may include an increased ability to recruit high threshold motor units (Ronnestad, 2009). As well as differences in overall muscular strength and the profiles of force-velocity characteristics. Training histories have been suggested to influence the response to WBV frequency (Ronnestad, 2009).

The present study also conflicts with Ritzmann *et al.* (2012) in which quadriceps muscle EMG activity was reported to be dependent on WBV frequency. The use of different EMG filtering techniques (band pass filter of 20 – 450 Hz versus 10 – 1000 Hz respectively) and different EMG data collection durations during WBV (2 x 10 s samples versus 1 x 10 s sample respectively) may account for difference in findings regarding the influence of WBV frequency on EMG activity. The WBV frequencies utilised were limited to 5 – 30 Hz (Galileo WBV) and it may be possible that at the higher frequencies (30 – 50 Hz) utilised in the present study the EMG response to frequency may be different. This could well be based on muscle tuning mechanisms (see section 2.3.6). Specifically, for the quadriceps muscles, motor unit discharge rates of approximately 50 impulses per second are associated with the greatest force generating capacity (Edwards et al., 1977). As discharge rates of most human spindles are directly proportional to WBV frequency; 50 Hz WBV may generate the 50 impulses per second associated with greater force generating capacity (Ronnestad, 2009). Therefore, the higher WBV frequencies utilised in the present study may influence the EMG response compared to lower frequencies utilised in previous literature. Finally, the present study conflicts with findings suggesting 30 Hz elicits the greatest fatigue response (Mischi et al., 2010). No one single frequency was found to be fatiguing; however, the work by Mischi *et al.* (2010) was based on superimposed vibration applied to the upper body, and not WBV.

The present study finding of no main effect for frequency across all participants includes 0 Hz in which they were volume-matched. Therefore, it could be concluded that 60 s of WBV at any of the four frequencies (30, 35, 40 and 50 Hz) does not appear to elicit a higher response in EMG activity. If taken at an individual level (see Figure 5.6) some participants elicited higher EMG responses at all four WBV frequencies compared to 0 Hz. However, the present findings of participants as a group conflict with Marin *et al.* (2009) who reported increases in vastus lateralis EMG activity during 30 Hz WBV versus no WBV. The likely reasons for this conflict are differences in: WBV platform type (iTONIC versus VibraMachine respectively); WBV duration (30 versus 60 s respectively); and D_{PTP} magnitude (8 versus 3 mm respectively). Differences in platform type may affect the EMG response to WBV as it is dependent on the direction of oscillation (Abercromby *et al.*, 2007b; Ritzmann *et al.*, 2012).

The present study also conflicts with Jordon *et al.* (2010) who reported a decrease in muscle activation following WBV. Even though, at individual level, some participants did show signs of reduced EMG activity; at group level no such significant decrease in EMG activity was reported. Again, the WBV oscillatory direction is a likely reason for the conflict as is the WBV duration; as Jordon *et al.* (2010) utilised a 3 x 60 s protocol compared to 1 x 60 s in this present study. The higher number of repetition may explain the decrease in EMG activity reported especially as a volume-matched control intervention also reported decreased EMG activity. However, it is important to highlight that the EMG data was recorded during an electrically evoked twitch response protocol; and not during WBV as completed by this present study.

Based on the theoretical mechanisms outlined in section 2.3.4 it may be expected that WBV would increase EMG activity. As already mentioned in section 4.4, WBV is likely to increase neural flow within a stretch reflex loop induced by changes in muscle length (Cardinale & Bosco, 2003; Jordan *et al.*, 2005). As a consequence: neuromuscular facilitation may occur involving increased synchronisation, recruitment and/or coordination (Cardinale & Lim, 2003b). It is perhaps surprising that this present study did not report any main effects for frequency across participants as a group. There are a number of possible reasons for this.

Firstly, the knee angle adopted has been reported to influence EMG response during varying WBV frequencies (Ritzmann et al., 2012). At higher knee flexion angles voluntary activation of quadriceps muscle groups is likely to offset higher joint torque generated (Ritzmann et al., 2012). Although this was only investigated in knee angles up to 60 º. Knee angle is likely to influence motor unit recruitment during isometric positions, and thus EMG activity (Pincivero et al., 2004). It has also been reported that at greater knee flexion angles the quadriceps muscle group is placed under an increased degree of stretch potentiating the effectiveness of WBV stimuli (Cardinale & Lim, 2003b). The present study utilised a posture of 90 º knee angle for that reason. Therefore, it seems likely the muscle group sampled for EMG activity (vastus lateralis and rectus femoris) would have been placed under stretch. Ritzmann *et al.* (2012) suggested that at higher knee flexion angles WBV frequency would have a greater influence on EMG activity. This was not found in the present study but does support Roelants *et al.* (2006) who reported no effect of knee angle on EMG activity.

Vastus lateralis and vastus medialis have displayed muscle specific responses to knee angle (Pincivero et al., 2004). By utilising 90 º knee angles under the aim of placing the quadriceps muscle group under stretch, in relation to WBV stimuli, the present study may not have chosen the optimal knee angle in relation to EMG activity during squatting. For both vastus lateralis and rectus femoris, 70 to 80 º knee angles have elicited the highest EMG activity response (Brownstein et al., 1985). Whereas, a 90º knee angle was ranked 5th for the same EMG response. This was based on non-weight bearing IMVC; nevertheless 90 º may not have been the optimal knee angle to elicit the highest EMG activity; and may attenuate a potential WBV frequency specific response.

Secondly, the filtering methods of EMG data collected during WBV may have also influenced the results. This aspect will be covered in chapter 6; but in brief, the presence of an electromagnetic field generated by the WBV platform may cause the inclusion of noise artifacts within EMG data. This would be relevant as the WBV frequency was varied in the present study, potentially varying the frequency characteristics of noise artifacts and sub harmonics. This may have influenced the overall results with regards EMG data as well as main effects for frequency.

Finally, the recruitment of trained participants is likely to have influenced the results. It has been suggested that well-trained individuals may have a smaller increase in EMG activity from WBV stimuli due to higher pre-existing muscle activation levels (Ronnestad et al., 2012). Previous literature has also failed to illustrate a beneficial response to WBV in well-trained individuals (Bullock et al., 2009). Albeit, the performance marker was sprinting and no EMG data was collected. However, changes in muscle-tendon complex characteristics may dampen WBV stimuli by acting as a mechanical buffer (Bullock et al., 2009). As a consequence the transmission of WBV to the quadriceps may be reduced in well-trained individuals. This aspect will be investigated in chapter 7. A higher magnitude of WBV stimuli may be required, supporting the argument that a D_{PTP} of 3 mm may not be sufficient to evoke a statistically significant EMG response, a point very pertinent to chapter 8.

5.4.2 The effect of whole body vibration frequency on overall jump performance

The finding of this present study that acute WBV, regardless of frequency, did not change overall jump performance across participants supports previous literature (Di Giminiani et al., 2009). The authors utilised an individualised WBV frequency (as determined by the highest EMG response) reporting no change in CMJ height. The findings also support Armstrong *et al.* (2010) who reported no effect of WBV frequency (30, 35, 40 and 50 Hz) on CMJ performance.

It does support Guggenheimer *et al.* (2009) who reported no significant frequency interaction on exercise performance. Although, the protocol utilised involved “high knee running” on the plate and the influence on WBV stimuli is unknown. The study also supports that of Gerodimos *et al.* (2010) who reported no influence of WBV frequency on jump performance, albeit SQJ. Finally, the results also support a meta-analysis review who reported acute vertical WBV did not benefit overall jump performance (Marin & Rhea, 2010a). Yet, the same authors admitted there is insufficient data to comment on specific variables and further research is warranted.

The findings of the present study conflict with those reported by Bedient *et al.* (2009) as main effects for frequency was found in CMJ performance; reporting improvements following 30, 35 and 40 Hz versus 50 Hz WBV. Reasons for this conflict are likely to stem from differences in: WBV platform type; knee angle adopted (50 ° knee flexion angle); and participants of both male and female being included; all of which has been previously discussed to influence WBV response (see: section 2.2.2; section 2.2.3.1; and section 2.2.3.3). The work by Adams *et al.* (2009) also conflicts this present study reporting a reciprocal relationship with frequency and D_{PTP} on jump performance. The present study utilised a range of frequencies but controlled for D_{PTP} remaining constant at 3 mm. Therefore the overall WBV stimuli may not have been sufficient to elicit a jump performance response. An important aspect regarding D_{PTP} will be discussed in chapter 8. Again, the use of different WBV platform types may account for the conflict.

The present study finding also conflicts with Da Silva-Grigoletto *et al.* (2011) who reported increases in CMJ height following WBV. The study utilised a similar WBV platform type (vertical) and WBV parameters (30 Hz and 4 mm D_{PTP}) as the present study; as well as standardising arm movements during CMJ performances. It seems surprising that the present study conflicts with that of Da Silva-Grigoletto *et al.* (2011), but the recruitment of active participants with no history of resistance training may explain this. For less trained active participants the WBV parameters used by Da Silva-Grigoletto *et al.* (2011) were sufficient to elicit a positive response. Similar parameters (such as one repetition of 60 s), when exposed to trained individuals, may not have been sufficient to elicit the same response. Given that three or six sets of 60 s had been found to be beneficial in CMJ performance in untrained participants (Da Silva-Grigoletto et al., 2011). This adds weight to the argument of diminished effects following acute WBV in trained individuals as discussed at the end of section 5.4.1 and previous literature (Bullock *et al.*, 2009; Ronnestad *et al.*, 2012). The rationale for selecting this duration was from literature utilising highly trained participants (Cardinale & Lim, 2003b). However, this protocol involved higher D_{PTP} (10 versus 3 mm). Therefore suggesting the stimulus was not sufficient for the trained participants of this present study.

Lamont *et al.* (2010a) demonstrates the conflict within WBV literature by both supporting and contradicting different aspects of the present study, as both PAP and post activation depression (PAD) responses were reported. Overall, at 7.5 minutes post WBV only 50 Hz elicited beneficial responses in CMJ performance. No such PAP response following 50 Hz was reported by the present study. One such reason may have been the timing of post jump tests following WBV exposure. Recovery time within PAP mechanisms have been suggested to be up to 8 to 12 minutes post pre-load (e.g. WBV) to elicit beneficial PAP responses in CMJ (Kilduff *et al.*, 2007; Kilduff *et al.*, 2008). Therefore, the timings of post WBV jump tests may explain the conflict of this present study. This supports previous literature reporting that for well-trained athletes the effect of vertical WBV may be less (Marin & Rhea, 2010b), and 60 s may not be sufficient to induce muscle co-activation (Sanudo et al., 2012). In trained participants 50 Hz did not influence CMJ performance, whereas in untrained 50 Hz WBV was beneficial (Ronnestad, 2009), suggesting a trained-specific response to frequency. It appears that for trained participants recruited in this present study, a higher WBV duration or repetition number may have elicited a frequency response.

The use of both 60 s WBV duration and 90 º knee angle may have had an inhibitory effect on motor unit recruitment (Bagheri et al., 2012). This would influence CMJ performance as the same study reported no change in CMJ RFD. The 60 s coupled with 90 º knee angle adopted may have added to the inhibitory effect that is proposed to act on the antagonist muscle (e.g. hamstrings) (Cardinale & Bosco, 2003). However, by the nature of WBV, both agonist and antagonist muscle groups (quadriceps and hamstrings) are exposed to WBV stimuli and therefore may add to the inhibitory effect (Bagheri et al., 2012). A higher knee flexion angle may have changed the agonist/antagonist dynamics affecting motor unit recruitment and therefore potentially jump performance post WBV.

Differences between the response of jump performance to frequency may depend on training status due to motor unit recruitment ability (Ronnestad, 2009). If untrained participants are likely to have a larger response to WBV due to a greater capacity to recruit high – threshold motor units; (Ronnestad, 2009) then the reverse could be said for well-trained. For these individuals training adaptations are likely to influence neuromuscular and tendo-muscular characteristics reducing the capacity to improve further due to WBV stimuli. During CMJ this higher pre-existing level of recruitment in trained participants may mean lesser subsequent improvements from WBV (Ronnestad, 2009). It may be speculated that trained individuals reach a ceiling effect, via diminished improvements following WBV compared to untrained. One of the inclusion criteria of this present study was at least 6 months of resistance training. It can, therefore, be reasonably assumed that all participants were trained and had achieved high pre-existing level of motor unit recruitment. In addition, participants had plyometric training experience therefore, diminishing potential WBV improvements in stretch reflex type activity (e.g. RSI).

As discussed in section 2.3.7 PAP may be a potential mechanism accounting for a WBV response. If so, the WBV stimuli could be characterised as a pre load activity. PAP literature often utilises dynamic pre load activities, whereas the present study pre load activity was isometric in nature. The timing of post WBV tests was based on post WBV responses peaking at 60 s (Da Silva-Grigoletto et al., 2011) and remaining significant within 5 minutes (Adams et al., 2009). With reference to PAP it appears that this approach may have under-estimated the recovery time required for a PAP response. Recent literature suggests the use of isometric activities may offer no benefit, or be detrimental, to PAP (Crewther et al., 2012). Therefore, the isometric posture adopted during WBV exposure seems unlikely to elicit a desired PAP effect, accounting for the overall lack of response to frequency.

The protocol aimed to balance assessing jump performance post WBV as soon as possible with adequate recovery between jump attempts and types. There is conflicting WBV literature suggesting that five minutes post WBV would still be inclusive of WBV responses; or time enough for a decline in response (Adams et al., 2009) *cf.* (Bedient et al., 2009). This suggests post WBV DJ tests may have been outside the ideal window to detect possible responses to WBV frequency. As 120 s is suggested as the most effective timeframe to assess post WBV response (Da Silva-Grigoletto et al., 2009), the present study protocol would have only assessed the first and second CMJ attempts and none of the DJ attempts. This may explain the lack of overall response to frequency on jump performance.

It is proposed that PAP succeeds a period of fatigue and is characterised by a timeframe of 3 to 8 minutes (Sale, 2002; Lamont *et al.*, 2010a). A vibration specific H-reflex depression lasting up to 100 s has been reported (Bove et al., 2003); which may characterise a specific post WBV PAP fatigue timeframe (Lamont et al., 2010a). Therefore, the post WBV protocol in the present study may not have allowed PAP to develop. The use of MVC for an EMG normalisation protocol may have affected participants' responses to WBV (Pellegrini et al., 2010). By completing a MVC protocol activation levels would have been high and it has been suggested that WBV response may be higher following sub maximal activity (Pellegrini et al., 2010). A final point regarding the timeframe of post WBV tests is that an individualised response to the intervals between WBV exposure and testing may exist and was variable across participants (Dabbs et al., 2011). If individual differences in optimal recovery exist, this may have affected the response of participants to standardised post WBV timings. Further discussion on individualised responses will be dealt with later in section 5.4.3.

The majority of WBV has measured either CMJ height or CMJ power using a range of measurement techniques (contact mats, linear encoder and force plate). One of the outcome measures utilised by this present study was RFD. The finding that frequency during acute WBV did not influence RFD across all participants as a group is perhaps not surprising as no other jump performance measure was significantly altered. The finding does support previous literature in which no significant difference in CMJ RFD was reported following a 90 º knee flexion posture adopted during acute tri-planar WBV (Bagheri et al., 2012). The same study also reported no significant change in maximum force developed during CMJ, supporting this present study. The use of RFD during a CMJ remains controversial as RFD is often utilised during an isometric activity; aiming to measure motor unit performance such as recruitment and speed characteristics (Young, 1995). The reliability (% CV) was poor in relation to other jump performance measures (see Table 5.1). The effect of acute WBV on RFD obtained via isometric activities has received little research attention. To date, one study reported no acute effects of tri-planar WBV on RFD obtained via electrically evoked isometric activity (Hannah et al., 2012). This again would support the findings of the present study; however the nature of isometric and dynamic activities in obtaining RFD with respect to the specificity of athletic dynamic performance should be highlighted. In addition the nature of typical RFD recordings is via electrically evoked motor unit recruitment which is different from voluntary recruitment.

The final jump performance outcome measure was RSI; again, this has received, to date, little specific WBV research. The present study reported no influence of WBV frequency on overall RSI across participants. This conflicts with Di Giminiani *et al.* (2009) who reported a similar WBV type increased continuous rebound JH, which is a likely measure of explosive capabilities characterised during fast shortening cycles (Flanagan et al., 2008). A possible reason for the conflict is due to the chronic nature of the WBV programme (8 weeks) utilised in Di Giminiani's work. It may be that for RSI in particular the current WBV parameters utilised in an acute study design were not sufficient to evoke a response.

The use of CMJ as an outcome measure may have influenced the response reported. Previous literature has failed to record a frequency-related response in CMJ performance (Di Giminiani et al., 2009). It has been suggested that CMJ kinematics consists of slower stretching speeds with larger magnitude of angular displacement (Young, 1995; Bosco *et al.*, 1998). It could be argued that CMJ may not evoke a stretch reflex and therefore may not be as susceptible to a frequency response (Di Giminiani et al., 2009). This is in comparison with faster stretching speeds and smaller angular displacement values such as those occurring during a DJ; which has seen a WBV response when no CMJ response was elicited. Therefore, it seems the influence of WBV frequency on jumps involving slower stretching speeds and larger angular displacements may be reduced (Di Giminiani et al., 2009) and may explain the lack of WBV response in CMJ performance.

Subsequently, it would be expected that RSI (as measured via DJ) would be influenced by WBV frequency; which was not the case in the present study. The biomechanical and muscle recruitment differences between CMJ and DJ may explain this. During DJ, the triceps surae play a more significant role, rather than the quadriceps which contribute a role towards CMJ performance (Kovacs et al., 1999). In the present study EMG recordings were taken of the quadriceps muscle and the posture adopted during WBV was chosen to target this group. Therefore, the posture may not have targeted the triceps surae groups as effectively, placing them under stretch. In fact, to minimise vibration transmission to the head and torso, participants were instructed to stand with weight distributed on the forefoot. This would have involved a degree of plantar flexion and thus reduce any stretch placed on the triceps surae group. Therefore, the vibration stimuli may not have targeted the muscle group which would contribute towards DJ performance. Further research should investigate calf EMG activity during WBV. In this present study it is unknown which body segments or muscle groups were targeted by adopting 90 º squat, and what WBV transmission characteristics exist. These aspects will be investigated in chapter 7.

The box height from which participants initiated the DJ has been reported to influence VGRF (Wallace et al., 2010) and JH (Young, 1995); both of which influence RSI (Flanagan et al., 2008). An optimal box height may be established by either identifying the maximum JH or the highest RSI obtained through incremental box heights. However, these produce significantly different values (≥ 10 cm) (Byrne et al., 2010). Due to time restraints this was not completed in the present study. If below optimal, decreased tendo-muscular stiffness and neuromuscular pre-activation may reduce performance; above optimal, increased stretch load and eccentric work along with decreases in concentric work, reduce the capacity to use elastic energy (Read & Cisar, 2001). Although, there is also evidence to suggest that DJ technique may play a greater role than optimal box height in determining performance (Walsh et al., 2004).

A WBV stimulus of 60 s may have reduced an ability to maintain and transfer eccentric forces prior to the concentric phase starting during the DJ; thus reducing the concentric generation capacity (Lamont et al., 2009). This would not only affect the contact time but also the jump height, having a direct influence on RSI; which would account for the present study findings. Although, 60 s of WBV has been reported to improve jump performance (Da Silva-Grigoletto et al., 2011), highlighting again the conflicting nature of WBV literature.

The final reason to explain the lack of overall response in jump performance to WBV frequency may involve the participants' BM. The mean BM of participants in this present study was 82.6 ± 9.0 kg, with maximum value of 101 kg. The influence of BM on WBV stimuli delivered from this particular WBV platform requires further research, as BM may influence the magnitude of WBV output (Marin & Rhea, 2010b). This will be discussed in detail in chapter 7. At this point it could be speculated that a higher BM may have reduced the magnitude of WBV stimuli exposed to participants. This may account for the lack of response of jump performance to WBV frequency.

There are general limitations associated with this present study. Firstly, the small number of participants recruited after extensive searches may mean a lack of statistical power and responses to WBV frequency would have been difficult to detect. Previous studies have detected significant differences with similar participants numbers (Cochrane et al., 2008b). The inclusion criteria of age, gender and resistance training history did restrict recruitment; however these aspects were areas warranting further research thus adding to the WBV literature, and are methodological strengths. For example, standardising for gender has not occurred in previous literature, even though a potential gender-specific response to WBV has been reported (Bazett-Jones et al., 2008b). However, as a consequence a low number of participants were recruited and the study may be underpowered to detect significant differences in the outcome measures chosen.

No EMG data was analysed during either CMJ or DJ. Although the present study did investigate the effect of frequency on EMG activity, this was during WBV and not during jump performance tests completed pre and post WBV. The use of EMG data during these jump tests may provide useful insight into the neuromuscular response of WBV frequency, and therefore would be an area for further research.

The final limitation associated with jump performance is that the present study utilised a vertical WBV platform and was acute in nature. Therefore, comparisons across WBV platforms, which deliver varying vibration characteristics, should be avoided; given that oscillatory – specific responses have been reported (Ritzmann et al., 2012). Further research should also investigate the effect of frequency on jump performance response in well-trained participants over a chronic WBV programme.

5.4.3 Individualised responses to whole body vibration

5.4.3.1 Electromyography

As mentioned in section 5.4.1, the present study illustrated a potential individualised response to WBV frequency in terms of EMG response (see Figure 5.6). This supports the previous work outlined in chapter 4, despite the difference between elite highly trained participants and those well-trained participants recruited in this present study. Those individualised responses to WBV frequency support previous literature (Di Giminiani *et al.*, 2009; Di Giminiani *et al.*, 2010). It appears that the EMG response of participants may be dependent on individual differences in muscle fibre characteristics such as fibre type and muscle spindle quantities and locations (Armstrong et al., 2008). As the present study included participants who were trained in rugby and volleyball, these characteristics are likely to be developed, accounting for the individual responses illustrated. Differences in training and subsequent: strength; physical characteristics (such as somatotype); neuromuscular recruitment profiles; and explosive performance capacity may well account for the different responses to varying WBV frequencies, as mentioned in section 4.4.

The possible underlying mechanisms to account for an individualised response to WBV frequency was not a direct objective of this present study. However, the results add to the emerging literature suggesting a potential individual response and further research is warranted to investigate potential mechanisms. This may include utilising physical characteristics, training status, strength and explosive performance capabilities, as well as neuromuscular indicators such as muscle biopsies.

5.4.3.2 Jump performance

The present study also illustrated evidence of an individualised response of jump performance to WBV frequency. When the change pre-post WBV at 0 Hz was compared to the best change of the WBV frequency which an individual achieved their highest jump; significant differences were reported for all outcome measures, except CMJ height. In addition, for all jump performance measures change at an individual level appeared highly variable. Individual differences in jump performance response have also been reported; however, this was in SQJ performance only (Colson et al., 2010). Past reports of large inter and intra variability in RFD following WBV also supports this present study (Lamont et al., 2010b). The finding of individualised responses in jump performance to frequency also supports previous literature which presented large individualised variability in other neuromuscular outcome measures such as H-reflex response (Armstrong *et al.*, 2008; Pollock *et al.*, 2012); and motor unit firing rate during WBV (Pollock et al., 2012).

The possible reasons to explain an individualised jump performance response to WBV frequency include those discussed in sections 4.4 and 5.4.3.1. Differences in individual dampening mechanisms may explain the variability especially as motor unit firing rates can be associated with the vibration cycle (Pollock et al., 2012). These are thought to be phase locked with WBV frequency (i.e. as WBV frequency increases, motor unit firing rates also increased); however, the timing of the phase lock exhibited individual variability possibly due to differences in muscle fibre characteristics, cross-sectional area and sensitivity to WBV frequency (Di Giminiani *et al.*, 2010; Pollock *et al.*, 2012). By utilising individualised response in H-reflex to WBV, other possible reasons to account for the high variability such as: gender; training specificity; and history have been discounted (Armstrong et al., 2008). Reasons that could account for the variability may well be the physiological characteristics of muscle spindles, or even muscular strength.

In other PAP research, individuals with higher strength responded better from a heavy pre load and benefited greater from PAP (Hodgson et al., 2005). As a link between fibre type, strength and PAP responsiveness has been demonstrated (Hamada et al., 2000), it would appear other fibre characteristics are likely to influence the response to WBV frequency. However, as yet little research has been completed directly measuring WBV response and fibre characteristics via muscle biopsies for example. Finally, the influence of BM may account for the individual variability in response to frequency. As already mentioned the effect of BM on the magnitude of WBV stimuli is unknown, but may have a direct association with jump performance response post WBV. This possible influence will be discussed in chapter 7.

Another possible reason for the individualised response of CMJ and EMG activity to WBV frequency could be familiarity of the tasks by participants. All participants were very familiar with CMJ performances, but the majority of participants were unfamiliar with WBV. Many commented on the initial unusual feeling and it may mean that the EMG activity recorded consisted of postural adjustment elements, which may not have occurred during the familiar CMJ performance.

5.4.4 The role of expectancy effect during acute whole body vibration

To date, this is the first study to directly investigate the role of expectancy effect, in relation to WBV. Previous studies have attempted to control for a possible placebo effect (Luo *et al.*, 2008; Luo *et al.*, 2009). However, by telling participants of a sham intervention it nullifies the attempt to control for a placebo effect. Previous studies have blinded participants to which WBV intervention they were receiving; however, separate protocols completed by only those participants receiving individualised WBV programmes appear to suggest potential for additional awareness (Di Giminiani et al., 2009). This knowledge may be perceived as an additional benefit and a role of expectancy effect may ensue (McClung & Collins, 2007). But also vice versa as participants may try less hard in performance tests following other interventions (i.e. control).

No role of expectancy effect was reported by this present study. Whether participants thought they were receiving their individualised WBV frequency or not, jump performance did not change. In other words, the participants' performances were consistent whether they thought they were receiving their individualised WBV frequency; the performance responses to WBV were not significantly different. Therefore, it appears, that for this type of acute vertical WBV, using this particular set of WBV parameters, that participants' psychological perception may not have influenced their response in jump performance.

One possible reason for this could involve participants realising the aim of this part of the present study. The WBV frequency was blinded to participants throughout both parts of the study. Out of the sessions involving different frequencies (30, 35, 40 and 50 Hz) 79 % of participants' guesses were incorrect. This was based on questioning at the end of each session to establish if blinding of the frequency had been successful. There was an inherent 1 in 4 chance of correctly guessing the WBV frequency equating to a 25 % correct guess rate. In the present study this was 21 %, suggesting blinding was effective. However, the frequency was not double blinded, and although careful not to react to participants guesses, influence from the investigator cannot be completely ruled out.

Another aspect which cannot be ruled out is the ability of the participant to learn during visits. After experiencing each of the four WBV frequencies, participants' perception of what frequency they were experiencing during the role of expectancy part of the study may have increased. Therefore, when told they were receiving another frequency, (but were in fact still receiving their individualised frequency); participants may have realised and in effect "bypassed" the role of expectancy aim of the study. Participants may have gained this increased perception ability through differences in pitches of sound the WBV platform produced at the various frequencies. To overcome this, ear defenders could have been given to participants. This would have reduced the sound, which would have made communication with the participant problematic assessing any problems during the novel WBV stimuli. It could be speculated that as participants' visits were separated by at least 48 hours, any perception of sound may be short-lived.

A final reason why a role of expectancy effect was not reported may be the lack of response in jump performance to WBV frequency reported in this chapter. As no response was reported it may be that for these particular WBV stimuli parameters role of expectancy may not play a significant role. If WBV parameters were changed and a significant jump performance response to frequency was detected; then a role of expectancy effect may be evident. Further research into this area is warranted, perhaps replicating a previous WBV protocol which elicited a positive jump performance response, to assess whether a role of expectancy effect is present.

5.4.5 Conclusions

WBV frequency did not appear to consistently influence overall EMG or jump performance responses across participants as a group. This may have been due to: the knee angle adopted during WBV; the EMG filtering technique (see chapter 6); the training status of recruited participants; the timing of post WBV jump tests;

targeting the quadriceps muscle group over other groups; and finally, the influence of BM on the magnitude of the WBV stimulus (see chapter 7).

There appears evidence of an individual response of EMG to WBV frequency which may be due to individuality in: genetics; muscle fibre characteristics; location / quantities of mechanoreceptors; and performance capabilities. Finally, there appears to be little or no role of expectancy effect during acute WBV on jump performance. While this has a potential impact on past and future WBV literature, this was based on a WBV protocol which did not elicit a positive jump performance response across the group of participants. It is also based specifically on one particular acute WBV stimulus and the study design should be replicated on WBV protocols which have elicited a positive response in performance; as well as using different platform types.

Future research should investigate: the influence of filtering technique on EMG data collected during WBV; the EMG responses of WBV frequency in calf muscles; and the possible influence of external load on WBV magnitude. These will be discussed in chapters 6 and 7 respectively.

Chapter 6: The influence of filter technique on quadriceps electromyography data recorded during whole body vibration

6.1 INTRODUCTION

EMG measurements have been utilised to identify potential mechanisms involved in adaptations of the neuromuscular response to such a stimulus. Early work using EMG measures focussed on a singular recording electrode method (monopolar) based on a positive and highly linear relationship between EMG amplitude and force, e.g. see review DeVries (1968). However, more recently susceptibilities such as muscle cross talk and electromagnetic noise contamination have caused EMG research to utilise different recording methods (Defreitas et al., 2012). Muscle cross talk can consist of EMG activity from distant muscles, whilst electromagnetic noise contamination can emanate from power cords and electric machinery (Enoka, 2002). One such electric machine would be a WBV platform, and initial EMG data collected from chapters 4 (Figure 4.3), would appear to identify the presence of electromagnetic noise once the WBV platform is switched on.

Well established strategies, which aim to minimise these effects, include the use of bipolar electrodes (Defreitas et al., 2012) and limiting the range of frequencies included in the EMG recording; i.e. filtering (Enoka, 2002). The aim of a filtering technique is to exclude EMG activity which is not relevant to the EMG muscle activity signal. The nature of EMG measurement is that recordings are likely to contain both true muscle signal and various noise components, many of which are unavoidable (De Luca et al., 2010). This can lead to errors during EMG data interpretation and noise components need to be identified and investigated before conclusions are made (Ritzmann et al., 2010). The presence of these noise components can result in over-estimation of EMG power and EMG_{rms} (Fratini et al., 2009a). Applying filters post EMG recording (digital filtering) can be problematic as the aim is to filter the maximum amount of noise without excluding portions of signal that contain true muscle signal (De Luca et al., 2010). Digital filters vary by the portion of frequency it excludes: low pass filters exclude frequencies above that stated; high pass filters excludes those portions of signal below the stated frequency; band pass filters allow only portions of signal between two stated frequencies to pass and excludes all other portions below or above the stated range (De Luca et al., 2010).

Recommendations regarding filter characteristics remain wide ranging and varied with no empirically-based filter specifications for limb EMG data (De Luca et al., 2010). Some general recommendations exist such as utilising a low pass filter of approximately 300 Hz to eliminate tissue noise (Cram & Kasman, 1998). A "preferred" band pass filter of 20 – 300 Hz would display the most accurate EMG signal recording during typical movement based activities (Cram & Kasman, 1998). The order and type of band pass filter is also a variable, with Butterworth and Bessel filter types the most commonly selected due to advantages combining roll rate and phase lag characteristics (Kamen & Gabriel, 2010). The roll rate represents the slope indicating how strictly the filter enforces the cut off frequency (e.g. at 20 Hz and at 300 Hz); which ideally should be a vertical cut off but, as this is not possible, involves a slope within the transition leading up to the filter cut off frequency (Kamen & Gabriel, 2010). The phase lag represents the time delay of the output following filtering (Kamen & Gabriel, 2010). The order of filter influences the sharpness of the cut-off employed during the filter (Winter, 1990). Compromise is required as a cut-off frequency set too high, while distorting the true muscle signal less, will allow more noise to pass than is preferable (Winter, 1990). However, a cut-off frequency set too low, while eliminating noise drastically, will distort true muscle signal; therefore a sharp cut-off (i.e. a high order of filter) is recommended (Winter, 1990).

Work by De Luca *et al.* (2010) suggested that the majority of lower limb true muscle signal is contained within 20 – 200 Hz and the majority of artificially induced motion artifact were contained within 0 – 20 Hz. The first empirical filter recommendation based on isometric lower limb EMG recording was the use of a Butterworth high pass filter of 20 Hz. These recommendations are likely to exclude general noise components experienced during everyday EMG recording. However, the use of

EMG in WBV research presents its own unique challenges (e.g. electromagnetic noise).

Very few studies have investigated the effect of WBV platform generated electromagnetic noise on EMG recording. Motion artifacts normally excluded by band filters are now not confined to below 20 Hz (Fratini et al., 2009b). Inclusion of these in EMG recording can lead to estimation errors, including over-estimation of the muscular activity elicited by WBV (Fratini *et al.*, 2009a; Fratini *et al.*, 2009b). However, the work by Fratini *et al.* (2009b) is limited by the choice of WBV frequencies as these did not include 30, 35, 40 or 50 Hz (common preset frequencies). In fact, power spectra calculated from EMG recordings during WBV suggested 29 % of EMG signal was associated vibration (Fratini et al., 2009a). This was based on the power spectra characteristics illustrating vibration induced artifacts at sharp peaks of the corresponding fundamental frequency. It was reported that a significant difference between unfiltered and filtered EMG signal was found with a 30 % reduction in EMG_{rms} (Fratini et al., 2009a). The authors utilised a series of sharp notch filters which involved filtering a narrow range of frequency centred at the fundamental WBV frequency and its sub-harmonics. For example, EMG data collected during 30 Hz WBV, would involve the following series of notch filters: 28.5 – 31.5; 58.5 – 61.5; 88.5 – 91.5; and 118.5 – 121.5 Hz and so on. Yet studies which utilised EMG as a measure of individualised response have not discussed these artifacts, nor utilised notch filters (Di Giminiani et al., 2010).

This form of notch filtering appears to solve the unique problem associated with WBV electromagnetic noise. However, Fratini *et al.* (2009a) acknowledged that as the WBV-induced motion artifacts fall within the standard EMG frequency range, even narrow notch filters of ± 1.5 Hz will likely exclude proportions of true muscle signal (Fratini et al., 2009a). Further limitations of the work included: the trained nature of participants; and the methodology of utilising electrodes placed on the patella to record motion artifacts magnitude. This limitation was highlighted by Ritzmann *et al.* (2010) who argued that the use of patella location is susceptible to cross talk from quadriceps muscle group. The authors also utilised artificially-induced stretch reflexes with an aim to determine the origins of the previously mentioned sharp artifacts in EMG signal power spectra. The conclusion reported that motion artifacts induced by WBV electromagnetic noise contributed very little (7 %) towards the overall EMG signal power, (Ritzmann et al., 2010). The authors went further and concluded that the EMG signal had a strong contribution of stretch reflex activity, and therefore, the use of notch filters would filter out true EMG muscle signal.

The findings by Ritzmann *et al.* (2010) are limited by the use of a Galileo platform, utilising only 5 to 30 Hz frequencies. Additionally, the posture adopted on the platform was not given; from illustrations it appears to be approximately 5 – 10 º knee flexion angle. This posture is unlikely to place the quadriceps group under a degree of stretch, potentially limiting the WBV influence on stretch reflexes (see literature review section 2.2.3.3). The findings that motion artifacts contribute little, and true muscle signal is dominant, within EMG recordings during WBV should be put into context. Only the soleus muscle was investigated for stretch reflex activity and within untrained participants, therefore the same cannot be presumed for other: muscle groups, such as quadriceps; populations, such as well-trained participants; and WBV platforms, such as those generating WBV stimuli from frequencies greater than 30 Hz.

Ritzmann *et al.* (2010) contradicts a number of WBV EMG studies which have chosen the notch filter technique, often without detailed justification (Cormie *et al.*, 2006; Abercromby *et al.*, 2007a; Fratini *et al.*, 2009a; Fratini *et al.*, 2009b; Marin *et al.*, 2011). In WBV research there is a wide range of filter techniques employed: a standard band pass filter (Bongiovanni & Hagbarth, 1990; Bosco *et al.*, 1999a; Cardinale & Lim, 2003b; Mileva *et al.*, 2006; Melnyk *et al.*, 2008; Marin *et al.*, 2009; Di Giminiani *et al.*, 2010); or in fact no digital filter (Luo *et al.*, 2005b; Luo *et al.*, 2008; Luo *et al.*, 2009); although it is worthwhile noting these three studies involved locally applied vibration and not WBV. A fourth filtering technique has been utilised involving spectral smoothing and removal of motion artifacts called spectrum interpolation (Mewett *et al.*, 2004; Mileva *et al.*, 2009; Mischi & Cardinale, 2009). In addition, other filtering techniques used by Ritzmann *et al.* (2012) employed an integrated and time normalised processing technique (iEMG). This iEMG during WBV was normalised to the corresponding position adopted without WBV. Sebik *et al.* (2013) employed a harsher band pass filter (80 – 500 Hz) then rectified the EMG signal to obtain information regarding motor unit discharge rates. Both Ritzmann *et al.* (2012) and Sebik *et al.* (2013) are provided for reference only within this thesis.

As yet the effect of these different filtering techniques on EMG data collected during WBV is relatively unknown with regards to whether portions of EMG data are removed post filter application. Therefore, research is needed to investigate the influence of filter technique on other WBV platform types involving more commonly-used higher frequencies of WBV. This research should investigate the effect of filter techniques (two commonly used techniques: band pass filter; and band pass filter with notch filters) on EMG data collected from trained participants during WBV. More specifically, a lack of research into the effect of filter on quadriceps EMG data should be addressed. Therefore, the objective of this study was to establish whether filter technique influences quadriceps EMG data collected during WBV by trained participants. The hypothesis was that filtering technique would significantly influence EMG data collected during WBV; and that a band pass filter technique, and that a band pass plus notch filter technique would remove significant portions of EMG data.

6.2 METHODS

6.2.1 Experimental approach

To investigate the influence of filtering technique on EMG data collected during WBV, an analytical study design was implemented. This utilised EMG data sets from chapter 5, part 1 (see section 5.2.1). Three different filtering techniques were individually applied to identical EMG data sets. Therefore, for the purpose of the following methodology, participants, procedures and equipment are described in general method and study, part 1 method sections and will not be repeated here. By utilising data sets collected during the previous study, the research was inclusive in ethical approval given for chapter 5. As both data analysis and statistical analysis were specific to this chapter, these are discussed as follows.

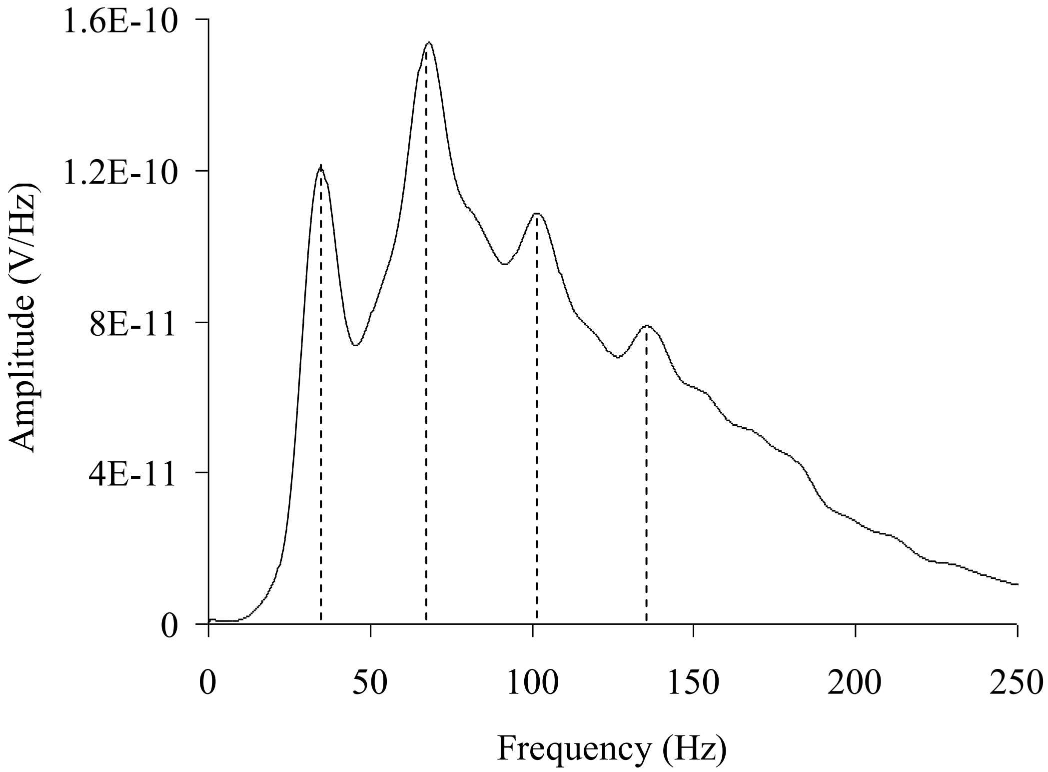
6.2.2 Data analysis

EMG data was taken from visits 1 to 5 for each participant and three different filter techniques were applied to the data (Unfiltered (no band pass filter or notch filter); Filter 1 (band pass filter); Filter 2 (band pass with additional notch filters). The resultant data was then averaged to allow analysis.

6.2.2.1 Fast Fourier Transformation (FFT)

FFT was performed for each WBV frequency (0, 30, 35, 40 and 50 Hz) across all 7 participants as explained in section 3.2.3. Power Spectral Density were determined, an example is given in Figure 6.1, illustrating peaks in power spectra at fundamental frequency (e.g. 35 Hz) and subsequent sub harmonics frequencies (e.g. 70, 105, and 140 Hz).

Figure 6.1: Power spectral density of EMG collected during 35 Hz WBV, illustrating peaks in power spectra at fundamental frequency (35 Hz) and relevant sub harmonics (70, 105 and 140 Hz).



6.2.2.2 Unfiltered

One of the three filtering techniques was to analyse unfiltered EMG_{rms} data. For clarification, the general method section outlined electrical filter techniques. Therefore, no digital filter was applied to EMG_{rms} data of the three muscles (Luo et al., 2008).

6.2.2.3 Filter 1

Using software (EMGworks Analysis, section 3.2.3) a 14th Order Butterworth band pass filter (20 – 300 Hz, passband 3 dB, attenuation 40 dB) was applied to EMG data of the three muscles recorded during 30, 35, 40 and 50 Hz WBV. EMG data was then root mean squared (EMG_{rms}) and transferred to spreadsheet for analysis.

6.2.2.4 Filter 2

Using the same software an identical band pass filter as filter 1 was applied to EMG data sets. In addition a series of 14th Order Butterworth notch filters (band stop ± 1.5 Hz centred at WBV fundamental frequency and relevant sub-harmonics, passband 3 dB, attenuation 40 dB) was applied across all 3 muscles (Abercromby *et al.*, 2007a; Fratini *et al.*, 2009a). For example, for EMG collected during 30 Hz, band stop filter centred at 28.5 – 31.5 Hz (fundamental frequency) and at relevant sub-harmonics: 58.5 – 61.5 (60 Hz); 88.5 – 91.5 (90 Hz); 118.5 – 121.5 (120 Hz); 148.5 – 151.5 (150 Hz); and 178.5 – 181.5 (180 Hz) (Fratini et al., 2009a). EMG data was then root mean squared as explained for filter 1.

6.2.2.5 Filter 2 versus 0 Hz "clean" electromyography data

"Clean" EMG data was identified as EMG recorded during 0Hz for each participant, i.e. the platform was not switched on. During the recording an identical posture was adopted by participants for an identical time period without any additional WBV stimulus. Therefore, the "clean" (0 Hz) EMG data was unlikely to contain motion artifacts at fundamental or at sub-harmonic frequencies. This is due to the lack of interference from the WBV platform when off. The same filter 2 process was applied to "clean" (0 Hz) EMG data to assess the impact of filter 2 in the removal of true muscle signal.

This process was completed for each frequency (fundamental and sub-harmonics) across all participants. For example, filter 2 applied to EMG data recorded at 40 Hz WBV would include notch filters ± 1.5 Hz band stop centred at 40, 80, 120, 160 and 240 Hz. Those same notch filters were applied to the corresponding participant's EMG data recorded during 0 Hz. This was completed for all three muscles and EMG data was then root means squared as explained for filter 1.

6.2.3 Statistical analysis

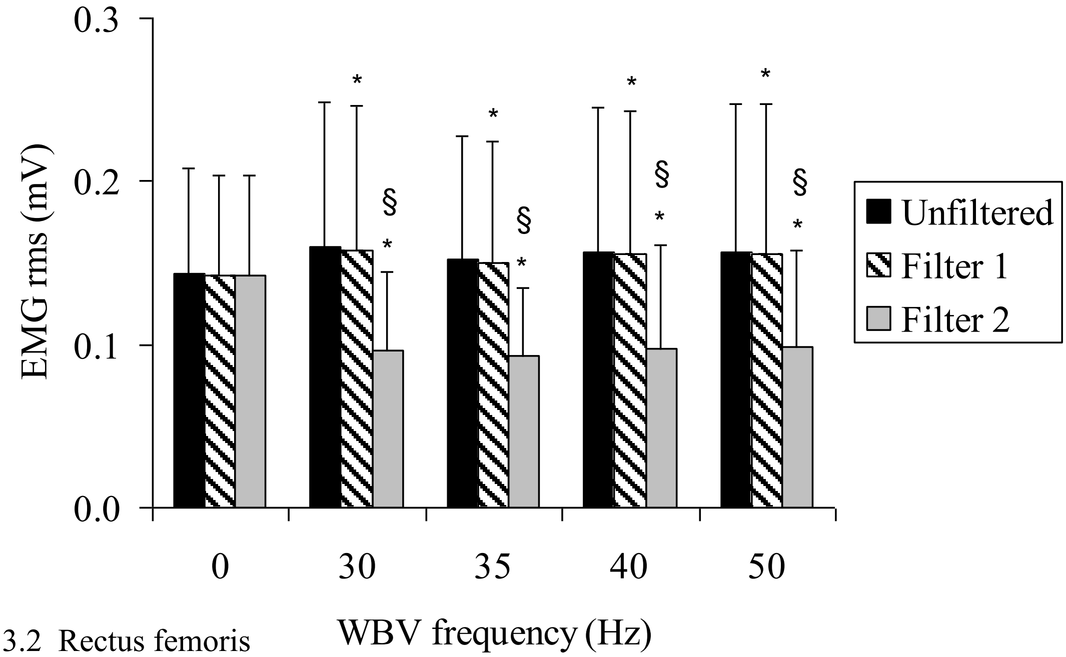
The effect of different filter techniques on EMG data recorded during different WBV frequencies was analysed by means of a 2-way repeated measures ANOVA (filter [unfiltered, filter 1, filter 2] x frequency [30, 35, 40, 50 Hz]). The difference between filter 2 applied to WBV (30, 35, 40 and 50 Hz) data sets and filter 2 applied to "clean" (0 Hz) EMG data was also analysed by means of a 2-way repeated measures ANOVA (filter [filter 2 applied to WBV EMG, filter 2 applied 0 Hz EMG] x frequency [30, 35, 40, 50 Hz]). Finally, two-tailed Pearson's correlation coefficients were calculated for vastus lateralis, rectus femoris and biceps femoris EMG_{rms} amplitude following the application of: unfiltered vs. filter 1; unfiltered vs. filter 2; and filter 1 vs. filter 2, for all WBV frequencies (0, 30, 35, 40 and 50 Hz). Coefficient of determination (r^2) was calculated and linear regressions for each WBV frequency were plotted along with true regression lines.

6.3 RESULTS

6.3.1 Vastus lateralis

A significant main effect for filter was found ($F_{[1.00,6.01]} = 38.98$, $p < 0.01$) with near perfect effect size ($r = 0.93$). Post hoc analysis revealed significant differences between: unfiltered and filter 1 (mean difference 0.002 mV, $p < 0.01$); unfiltered and filter 2 (mean difference 0.06 mV, $p < 0.01$); filter 1 and filter 2 (mean difference 0.058 mV, $p < 0.01$), Figure 6.2. No significant main effect for frequency was found ($F_{[1.49,8.96]} = 0.02$, $p = 0.96$) with small effect size ($r = 0.05$). No significant frequency x filter interaction was found ($F_{[1.55,9.32]} = 0.06$, $p = 0.91$) with small effect size ($r = 0.09$).

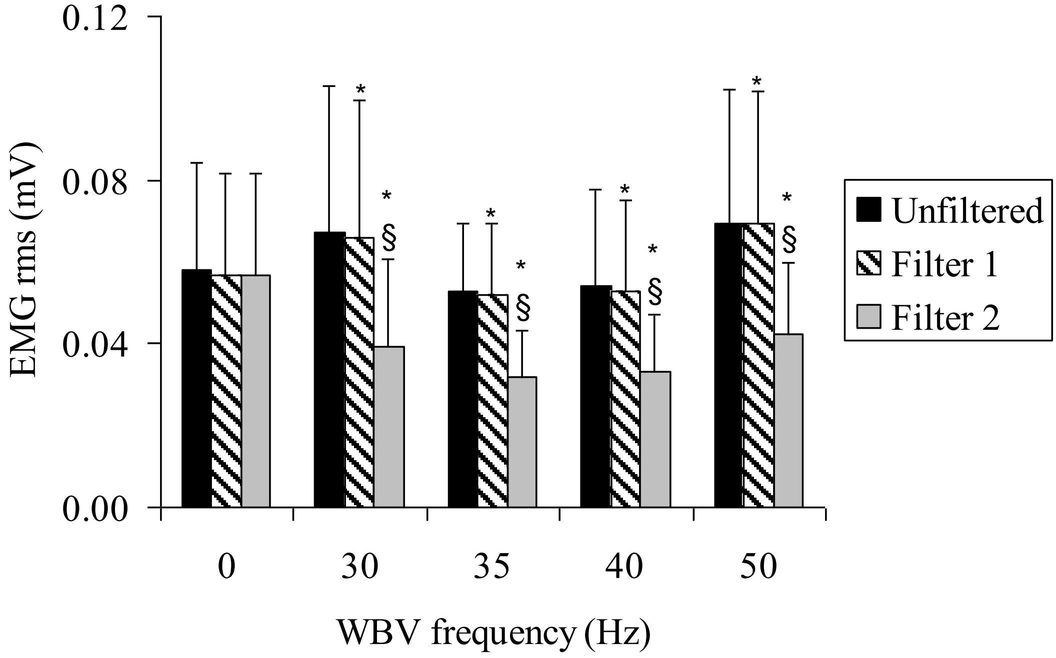
Figure 6.2: Mean ± SD vastus lateralis EMG_{rms} following three different filter techniques (unfiltered, filter 1 and filter 2) during several WBV frequencies. A significant main effect for filter was found. * Significantly lower than unfiltered, § significantly lower than filter 1 at the same WBV frequency.



6.3.2 Rectus femoris

A significant main effect for filter was found ($F_{[1.00,6.05]} = 67.20$, $p < 0.001$) with near perfect effect size ($r = 0.96$). Post hoc analysis revealed significant differences between: unfiltered and filter 1 (mean difference 0.001 mV, $p < 0.05$); unfiltered and filter 2 (mean difference 0.024 mV, $p < 0.01$); filter 1 and filter 2 (mean difference 0.023 mV, $p < 0.01$), Figure 6.3. No significant main effect for frequency was found ($F_{[3,18]} = 0.97$, $p = 0.43$) with moderate effect size ($r = 0.37$). No significant frequency x filter interaction was found ($F_{[1.56,9.35]} = 1.01$, $p = 0.38$) with moderate effect size ($r = 0.37$).

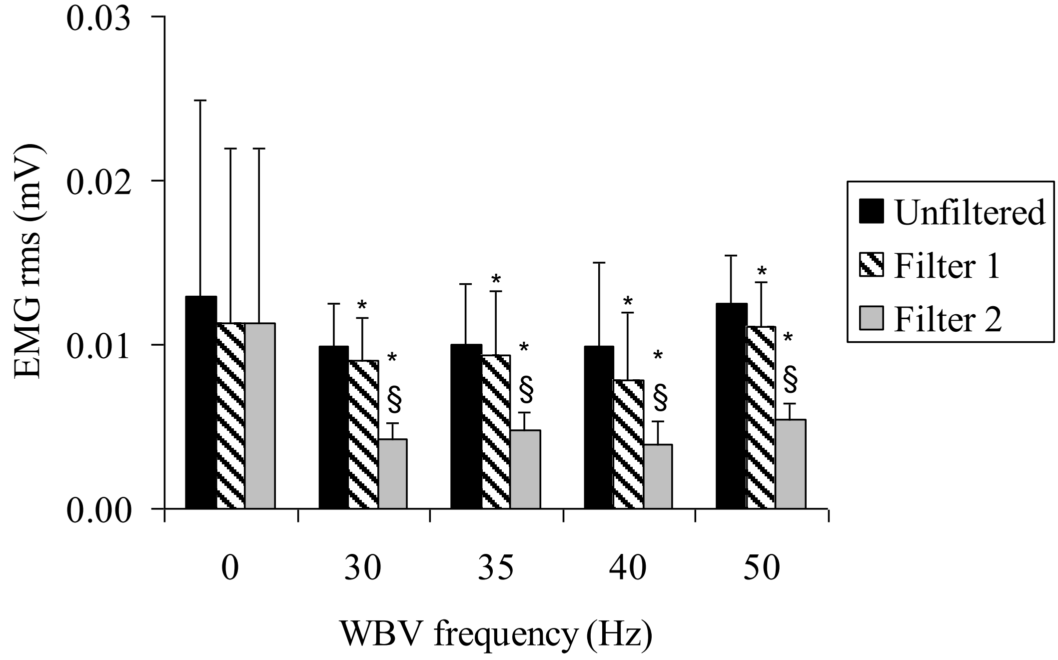
Figure 6.3: Mean ± SD rectus femoris EMG_{rms} following three different filter techniques (unfiltered, filter 1 and filter 2) during several WBV frequencies. A significant main effect for filter was found. * Significantly lower than unfiltered, § significantly lower than filter 1 at the same WBV frequency.



6.3.3 Biceps femoris

A significant main effect for filter was found (F [2,12] = 103.01, $p < 0.001$) with near perfect effect size ($r = 0.97$). Post hoc analysis revealed significant differences between: unfiltered and filter 1 (mean difference 0.001 mV, $p < 0.01$); unfiltered and filter 2 (mean difference 0.006 mV, $p < 0.001$); filter 1 and filter 2 (mean difference 0.005 mV, $p < 0.001$), Figure 6.4. No significant main effect for frequency was found (F [3,18] = 1.30, $p = 0.31$ with moderate effect size ($r = 0.42$). No significant frequency x filter interaction was found (F [2.24,13.43] = 0.54, $p = 0.77$) with small effect size ($r = 0.28$).

Figure 6.4: Mean ± SD bicep femoris EMG_{rms} following three different filter techniques (unfiltered, filter 1 and filter 2) during several WBV frequencies. A significant main effect for filter was found. * Significantly lower than unfiltered, § significantly lower than filter 1 at the same WBV frequency.



6.3.4 Filter 2 versus 0 Hz "clean" electromyography data

No significant main effect for filter was found for vastus lateralis, rectus femoris and biceps femoris, (range: F = 0.26 – 0.38, p = 0.56 – 0.63) with small effect sizes (range: r = 0.20 – 0.24). No significant main effect for frequency was found for vastus lateralis and rectus femoris (range: F = 0.07 – 0.64, p = 0.60 – 0.90) with moderate effect sizes (range: r = 0.32 – 0.35). No significant filter x frequency interaction effects were found for vastus lateralis or rectus femoris (range: F = 0.01 – 1.09, p = 0.38 – 0.98) with small to moderate effect sizes (range: r = 0.03 – 0.39). A significant main effect for frequency and significant filter x frequency interaction was found for biceps femoris (range: F = 3.17 – 5.64, $p < 0.05$) with a moderate effect size (range: $r = 0.59 – 0.70$), although post hoc analysis revealed no significant differences.

6.3.5 Filter correlations

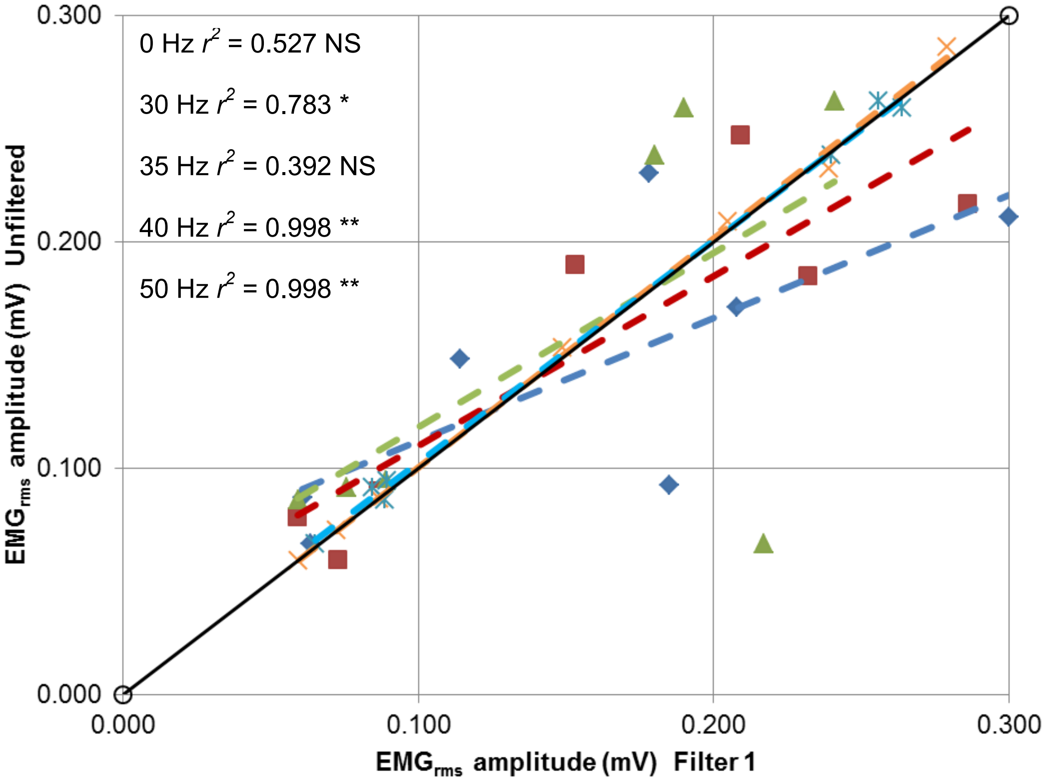
The relationship between unfiltered vs. filter 1 applications to vastus lateralis EMG_{rms} amplitude data appeared dependent upon WBV frequency (Figure 6.5, A). During two WBV frequencies (0 and 35 Hz) there were weak, non-significant relationships between unfiltered vs. filter 1 ($r^2 = 0.527$, $p = 0.07$; $r^2 = 0.392$, $p = 0.13$ respectively). Whereas, unfiltered EMG_{rms} amplitude data was significantly correlated with filter 1 data during other WBV frequencies (30, 40 and 50 Hz) ($r^2 = 0.783$, $p < 0.01$; $r^2 = 0.998$, $p < 0.001$; and $r^2 = 0.998$, $p < 0.001$ respectively).

Unfiltered vs. filter 2 vastus lateralis EMG_{rms} amplitude data was significantly correlated during all WBV frequencies ($r^2 \geq 0.978$, $p < 0.001$) (Figure 6.5, B). Filter 1 vs. filter 2 vastus lateralis EMG_{rms} amplitude data was significantly correlated during all WBV frequencies ($r^2 \geq 0.980$, $p < 0.001$) (Figure 6.5, C).

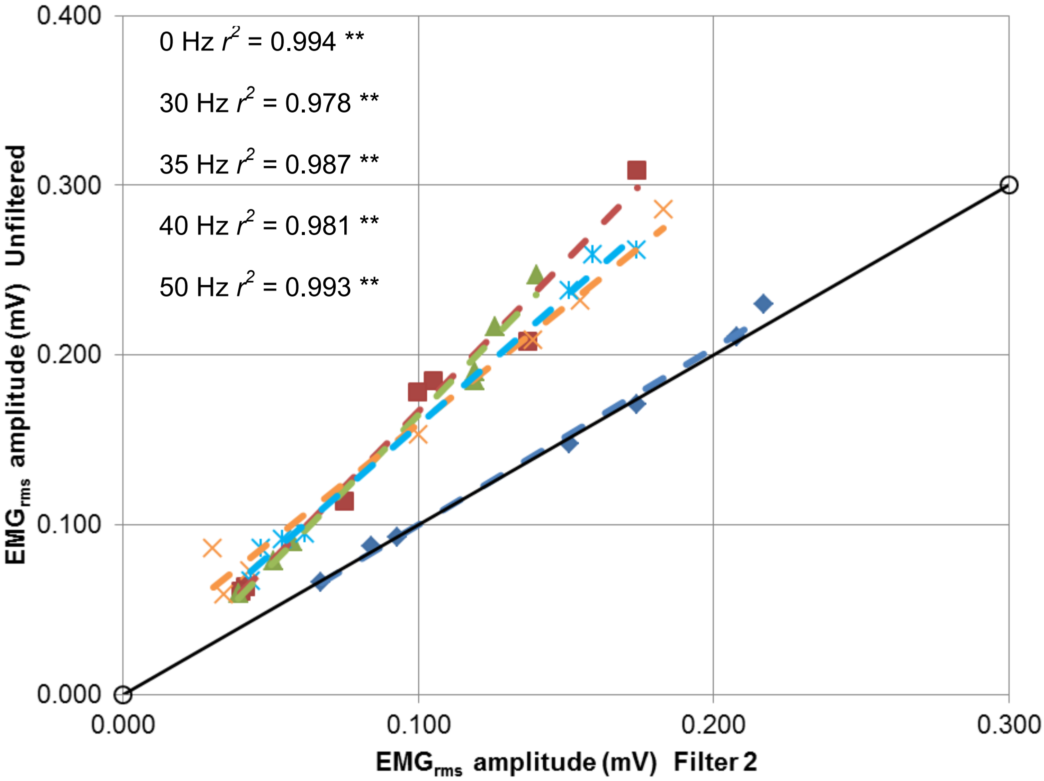
Similar correlation relationships were evident in rectus femoris EMG_{rms} amplitude data when comparing unfiltered vs. filter 1; unfiltered vs. filter 2; and filter 1 vs. filter 2 (Figure 6.6)

Figure 6.5: Vastus lateralis EMG_{rms} amplitude data correlations following the application of: A, unfiltered vs. filter 1; B, unfiltered vs. filter 2; C, filter 1 vs filter 2 across all WBV frequencies (r^2, coefficient of determination; NS, non-significant; *, $p < 0.01$; **, $p < 0.001$). Blue, 0 Hz; red, 30 Hz; green, 35 Hz; orange, 40 Hz; light blue, 50 Hz; black, true regression line.

A



B



C

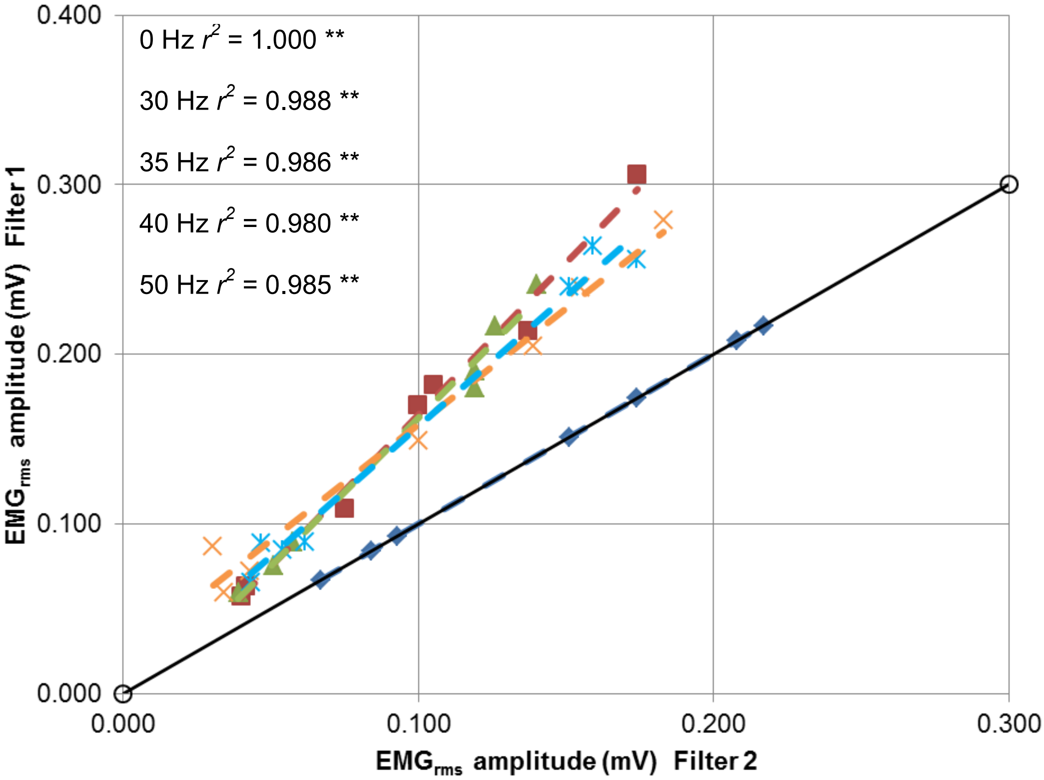
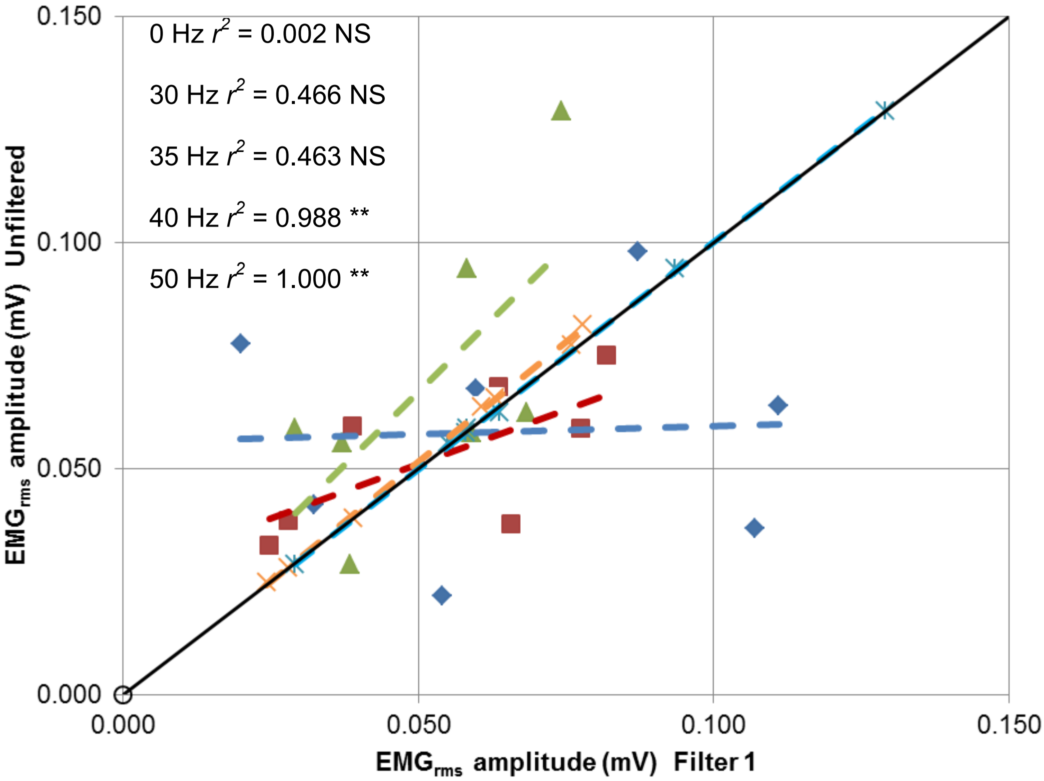
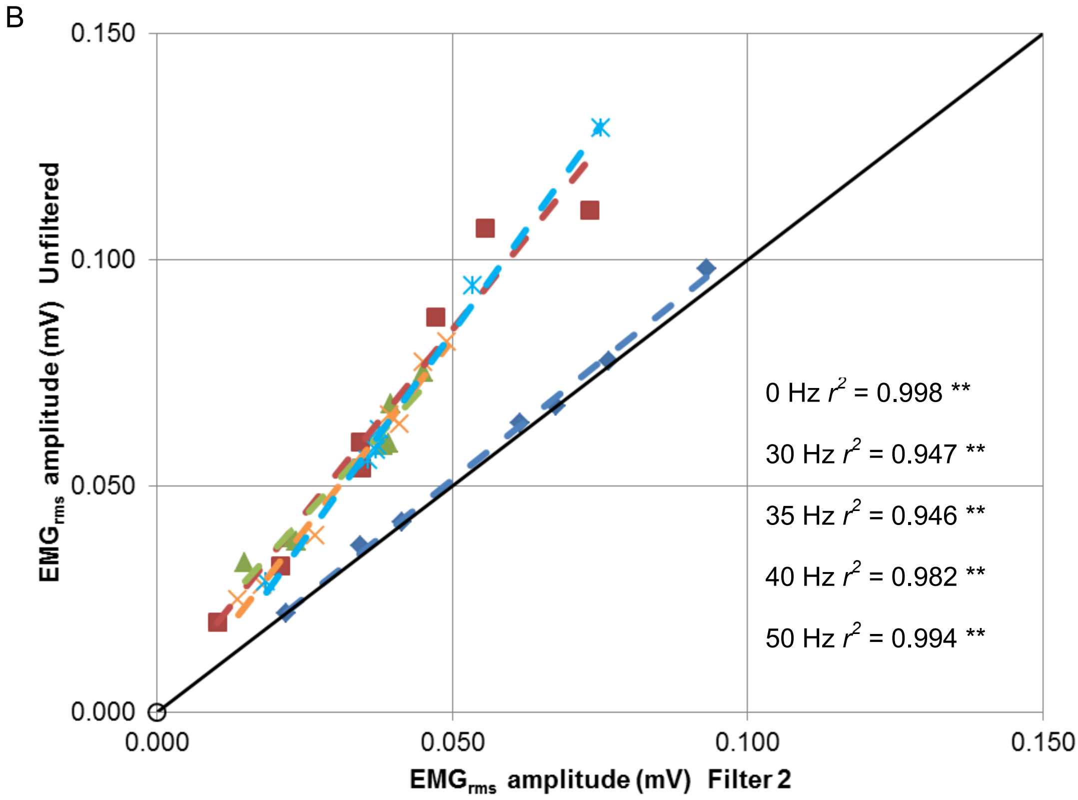
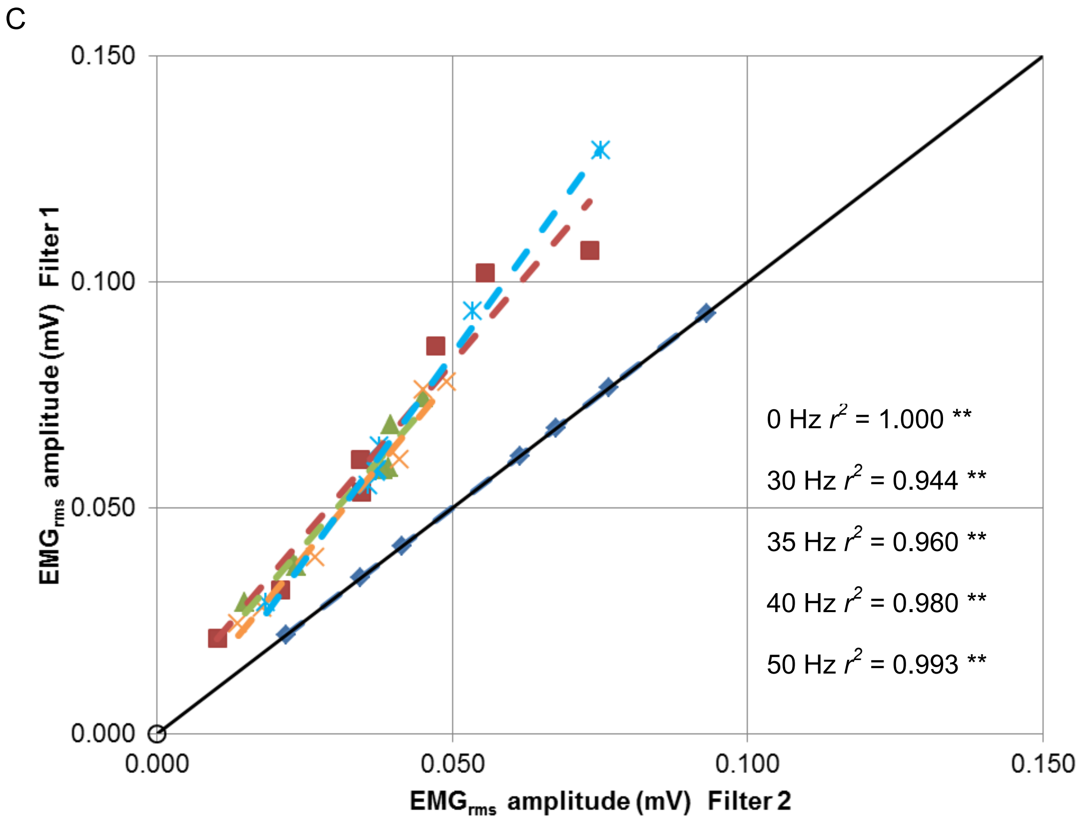


Figure 6.6: Rectus femoris EMG_{rms} amplitude data correlations following the application of: A, unfiltered vs. filter 1; B, unfiltered vs. filter 2; C, filter 1 vs filter 2 across all WBV frequencies (r^2, coefficient of determination; NS, non-significant; *, $p < 0.01$; **, $p < 0.001$). Blue, 0 Hz; red, 30 Hz; green, 35 Hz; orange, 40 Hz; light blue, 50 Hz; black, true regression line.

A




B
0.150
0.100
0.050
0.000
EMGrms amplitude (mV) Unfiltered
0.000
0.050
0.100
0.150
EMGrms amplitude (mV) Filter 2
0 Hz r2 = 0.998 **
30 Hz r2 = 0.947 **
35 Hz r2 = 0.946 **
40 Hz r2 = 0.982 **
50 Hz r2 = 0.994 **

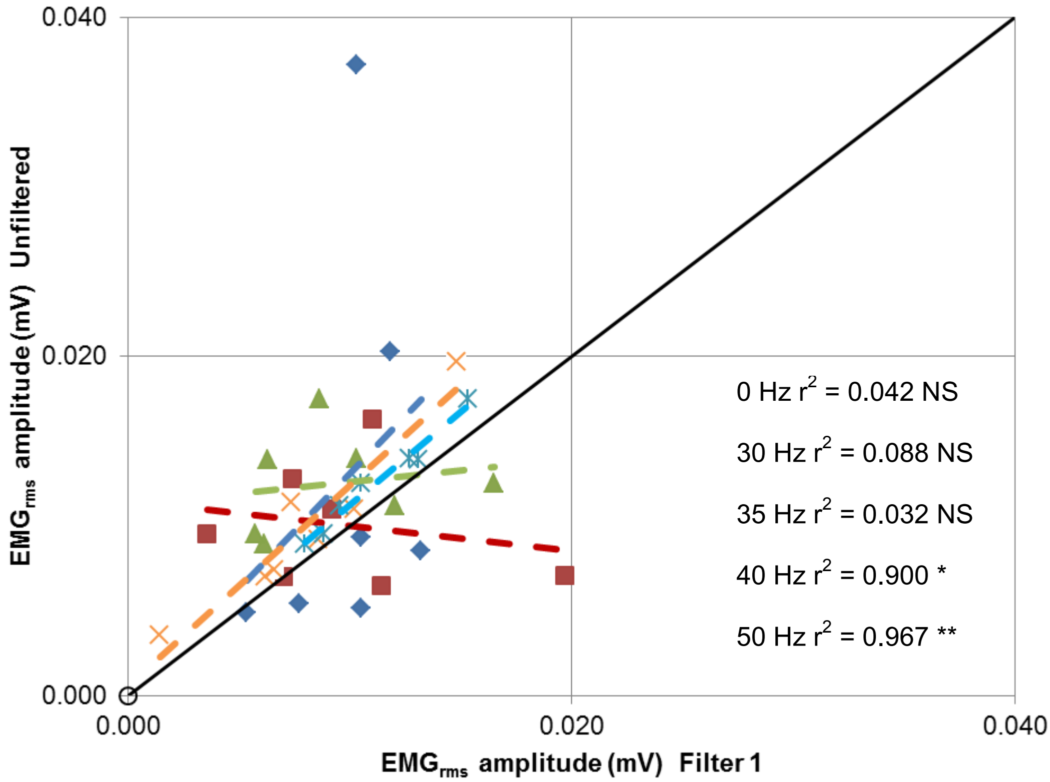

C
0.150
0.100
0.050
0.000
EMGrms amplitude (mV) Filter 1
0.000
0.050
0.100
0.150
EMGrms amplitude (mV) Filter 2
0 Hz r2 = 1.000 **
30 Hz r2 = 0.944 **
35 Hz r2 = 0.960 **
40 Hz r2 = 0.980 **
50 Hz r2 = 0.993 **

Unfiltered vs. filter 1 biceps femoris EMG_{rms} amplitude data followed a similar pattern to that of vastus lateralis, as significant relationships appeared dependent upon WBV frequency (Figure 6.7, A). During three WBV frequencies (0, 30 and 35 Hz) there were weak, non-significant relationships between unfiltered vs. filter 1 (r^2 = 0.042, p = 0.61; r^2 = 0.088, p = 0.57; and r^2 = 0.032, p = 0.69 respectively); whereas significant relationships during 40 and 50 Hz ($r^2 \geq 0.900$, $p < 0.01$) were found.

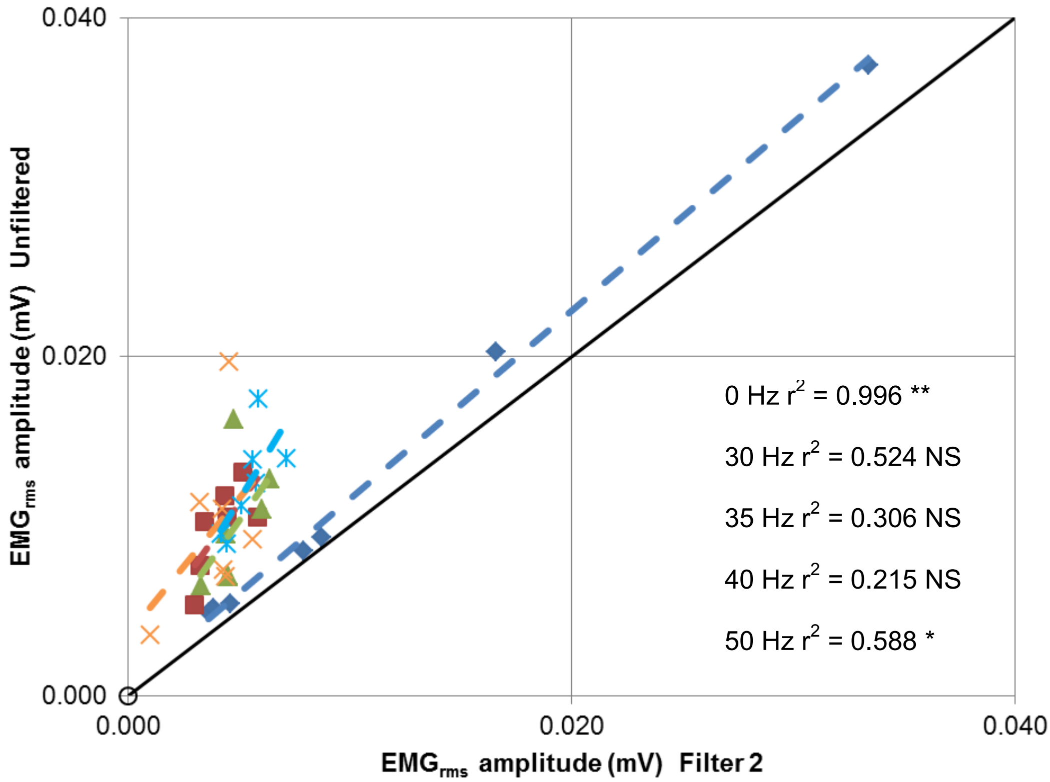
Unfiltered biceps femoris EMG_{rms} amplitude data was only significantly correlated with filter 2 data during 0 and 50 Hz frequencies (Figure 6.7, B). Biceps femoris EMG_{rms} amplitude data was only significantly correlated filter 1 vs. filter 2 during 0, 40 and 50 Hz WBV (Figure 6.7, C).

Figure 6.7: Biceps femoris EMG_{rms} amplitude data correlations following the application of: A, unfiltered vs. filter 1; B, unfiltered vs. filter 2; C, filter 1 vs filter 2 across all WBV frequencies (r^2, coefficient of determination; NS, non-significant; *, p < 0.01; **, p < 0.001). Blue, 0 Hz; red, 30 Hz; green, 35 Hz; orange, 40 Hz; light blue, 50 Hz; black, true regression line.

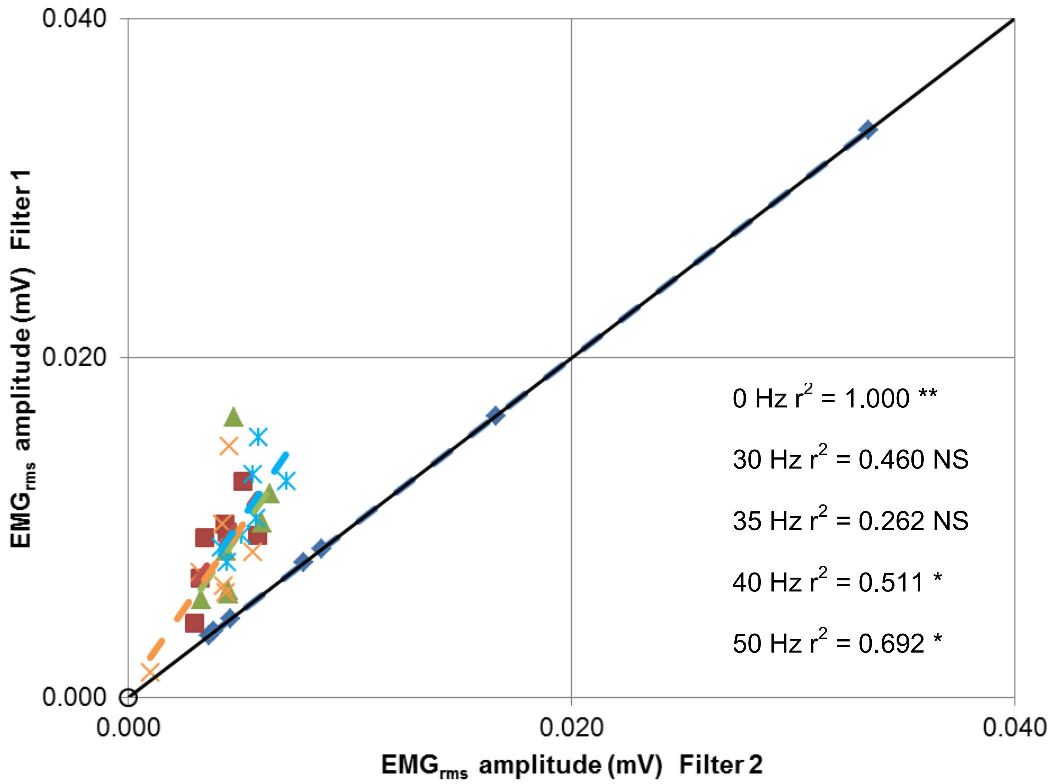
A



B


0.040
0.020
0.000
EMG$_{rms}$ amplitude (mV) Unfiltered
0.000
0.020
0.040
EMG$_{rms}$ amplitude (mV) Filter 2
0 Hz r^2 = 0.996 **
30 Hz r^2 = 0.524 NS
35 Hz r^2 = 0.306 NS
40 Hz r^2 = 0.215 NS
50 Hz r^2 = 0.588 *

C


0.040
0.020
0.000
EMG$_{rms}$ amplitude (mV) Filter 1
0.000
0.020
0.040
EMG$_{rms}$ amplitude (mV) Filter 2
0 Hz r^2 = 1.000 **
30 Hz r^2 = 0.460 NS
35 Hz r^2 = 0.262 NS
40 Hz r^2 = 0.511 *
50 Hz r^2 = 0.692 *

6.4 DISCUSSION

The main finding of this present study is that filtering techniques can have a significant influence on EMG data recorded during WBV from trained individuals. The results suggest that significant portions of EMG data signal were removed depending on which type of filter technique was utilised (notch filters > band pass filters > no filters). This was true for all three muscles recorded (vastus lateralis, rectus femoris and biceps femoris) during WBV of varying frequencies. The results also suggest that, for all three muscles, regardless of WBV frequency, applying notch type filters may not remove significant portions of EMG muscle activity signal. These results indicate that even though filter 2 may remove significant portions of EMG signal, it appears that it is not muscle signal. It also appears that for EMG_{rms} amplitude data measurements following the application of band pass filters versus no filter are different and dependent on WBV frequency. Whereas, when notch filtering is applied, there appears a uniformed reduction effect on EMG_{rms} amplitude.

The results support the overall approach advocated by Abercromby *et al.* (2007a) in which the potential for misinterpretation of EMG activity, especially during WBV, was put forward. The authors admitted even with existing strategies for reducing noise artifacts, such as securing cables and applying band pass filters, that over-estimation of EMG signal recorded during WBV can occur. The study also supported the findings of this present study with regards the presence of peaks in the power spectra centred at the excitation WBV frequency and its sub-harmonics (see Figure 6.1), attributing it to vibration of the EMG electrodes (Abercromby et al., 2007a).

The results also support both studies by Fratini *et al.* (2009a; 2009b) who reported that, by including these artifacts, muscle activity can be over-estimated. The presence of these artifacts is not confined to the power spectra range which band filter techniques target. Therefore, filtering techniques are likely to have significant influence over EMG data recorded during WBV (Fratini et al., 2009a), as found by this present study. The authors reported lower than normal D_{PTP} magnitudes (1.2

mm) during WBV and only for healthy, active participants. Therefore, the present study adds to the WBV literature investigating the influence of filtering techniques on WBV protocols with more commonly used D_{PTP} values and in more well-trained populations. The results also support the recommendation that a minimum of a band pass filter should be utilised, excluding frequencies below 20 Hz in the power spectra and that filter techniques should be carefully considered (De Luca et al., 2010). In addition, the relationship between EMG_{rms} amplitude data following the application of band pass filters versus unfiltered techniques may result in different post filter data. More importantly, this difference post filter application appears WBV frequency dependent as varying strengths of correlations were detected at the frequency range used in this present study. This may be related to a proportional relationship between WBV frequency and the degree of electromagnetic noise generated by the WBV platform; however, this area requires further research.

The work by De Luca and co workers conflicts with the present study as the majority of noise was reported to be contained within 0 to 20 Hz power spectra frequencies. This was not the case when analysing power spectra frequencies of EMG data collected during WBV, as reported by this present study. The recommendation of utilising a high pass filter of 20 Hz for isometric activity is not completely transferable to WBV use. The present study also contradicts the approach by Hazell *et al.* (2007) in which a wider band pass filter was utilised (100 – 450 Hz). Although sampling a similar range of WBV frequencies during EMG data collection; the use of this wider range of band pass filter was not compared to other filter techniques. This is especially relevant as excluding power spectra frequencies of 20 to 100 Hz may contain true muscle signal (De Luca et al., 2010).

Finally, the present study findings conflict with Ritzmann *et al.* (2010) who reported that contributions of noise artifacts to EMG signal recorded during WBV appear minimal. The authors speculated that, by applying filter techniques EMG data would be significantly altered, although no direct investigation to determine this was completed. The reason for the conflict with the present study may lie in the type of WBV frequencies used. Ritzmann *et al.* (2010) investigated Galileo WBV and as a consequence the findings are for pivotal WBV and only for frequencies between 5 and 30 Hz. The higher WBV frequencies used in this present study are likely to have a different effect on artifacts within the EMG recordings. As the fundamental WBV frequency will be lower and potentially within the power spectra frequencies that is routinely excluded by band pass filters (0 – 20 Hz). The type of oscillation may also explain the conflict as vertical WBV may exert different motion artifacts during EMG recordings.

As mentioned in section 6.1 there is a wide variety of filter techniques utilised by past WBV literature. More often than not, the choice of filter was not justified or rationale provided. By the nature of WBV platforms the choice of filter may prove crucial in correctly analysing EMG responses. The results of this present study have shown that performing different filter techniques significantly alters EMG data. Therefore, the choice of filter technique utilised may prove crucial. What is less clear at this moment is the content of the excluded EMG signal via filter 2 technique which has been shown by this present study to remove a significant portion of EMG. This portion is likely to have contained noise artifacts, but it may have also contained "true" muscle signal. This is due to the fact that the notch filters contained within filter 2 exclude EMG signal within the range of power spectra frequencies in which muscle activity signal lies (20 – 200 Hz) (De Luca et al., 2010). Therefore, at this point the results of the present study suggest the impact of filter technique on EMG data is significant. However, it cannot be argued that one filter technique is more effective than another (e.g. filter 2 versus filter 1); due to the fact it is unclear what EMG signal is being excluded by each filter.

To attempt to determine what signal type (noise or "true") is excluded via filter 2 technique a novel approach was devised. The study design outlined in chapter 5 included a control intervention (0 Hz); therefore participants adopted the same posture for an identical period of time, during which no WBV was initiated. As a consequence the EMG data recorded can be considered "clean" from electromagnetic noise produced when the WBV platform was on, (see previous FFT analysis Figure

6.1). Therefore, applying the same filter 2 technique (notch filters) to this "clean" EMG data set would likely only exclude signal that is "true" muscle activity.

By applying every notch filter (corresponding to each of the four fundamental WBV frequencies and their sub harmonics) to the "clean" (0 Hz) EMG data, there are two likely outcomes. If the subsequent EMG data were significantly different then it can be referred that notch filters may remove valuable "true" muscle signal. As no significant difference was reported for any of the specific notch filter it appears that by using these filter types that portions of "true" muscle signal may not be removed. In addition, the relationship between EMG_{rms} amplitudes following notch filters seem to suggest that the data signal is not a different measurement, merely a reduced magnitude following the filter application.

This is in conflict with Abercromby *et al.* (2007a) who suggested that the application of notch filter would exclude large portions of muscle signal within EMG data. As a consequence, this would result in an underestimation of the WBV response. However, the authors did not directly compare filter techniques as in this present study. In addition, the use of different WBV platform types highlights the potential difference in electromagnetic fields and therefore platform specific noise artifacts.

However, the novel approach of comparing filtered EMG data recorded during WBV with "clean" (0 Hz) EMG data has a potential limitation. By utilising the squat posture during 0 Hz the demands of adopting that posture have been accounted for. When attempting to account for the type of signal excluded during filtering; the additional demands of the WBV stimuli (eliciting rapid small changes in muscle length) has not been accounted for with this approach. The true muscle signal is likely to be different during WBV versus 0 Hz. Further research is warranted applying this approach as well as the filter technique itself to EMG data recorded from other types of WBV stimuli.

The results of the present study suggest that for acute vertical WBV delivered by this particular platform type, notch filter techniques effectively exclude noise artifacts without excluding significant portion of "true" muscle signal. Therefore, it is this study's recommendation that notch filtering should be utilised in further WBV research. As EMG data collected during WBV without filtering is likely to over-estimate the response to the stimuli. This is actually in agreement with Abercromby *et al.* (2007a) who commented that if notch filters were not applied, inaccurate EMG data analysis may occur.

The findings of this present study should be placed in context. The findings are based on the specific electromagnetic noise generated by this platform and may not be transferable to other platform types. The findings are also based on EMG data of trained individuals. Again, caution should be taken when applying these findings to other populations. Even by utilising an effective filter, the very process makes interpretation of the EMG data response to WBV difficult (Pollock et al., 2010). A balance is required as the use of no filter would mean no "true" muscle signal would be excluded; however, all noise artifacts would also be included. At the other end of the balance by applying an extensive filter all noise artifacts may be excluded but at the expense of "true" muscle signal. The use of a Faraday cage designed to exclude electromagnetic noise could be an alternative option. However, evidence suggests filtering is more effective in terms of signal to noise ratios (Defreitas et al., 2012).

6.4.1 Conclusions

In conclusion, filter technique significantly influences EMG data recorded during WBV. Previous literatures which have not used any filter technique may have over-estimated the EMG response to WBV due to specific platform related noise artifacts, which are contained within recorded EMG signal. The presence of these was confirmed and supports previous literature in identifying specific WBV related noise artifacts. Literature which has employed band pass filter or notch type filter techniques are likely to have significantly altered the EMG data sets. More consideration and justification is required with regards choice of filter technique as demonstrated in the results of this chapter.

This chapter attempted to characterise the excluded portions of EMG signal and suggested that it may not contain “true” muscle signal. However, by the very nature of WBV, muscle EMG activity during the stimulus is likely to be different than a matched posture during 0 Hz. These findings are also platform specific and recorded in trained individuals during WBV frequencies of 30, 35, 40 and 50 Hz only.

Finally, regardless of filter technique the results of chapter 5 still stand, as there was no frequency effect and furthermore, filtering technique effect was not frequency-dependent (and no interaction reported). Therefore filtering technique is unlikely to account for the findings of chapter 5. One possible explanation could be the influence of external load (see section 5.4.2) reducing the magnitude of WBV stimuli exposed to participants (see chapter 7).

Chapter 7: The influence of external load on electromyography and acceleration responses during whole body vibration at different frequencies

7.1 INTRODUCTION

One of the main findings in chapter 5 was that mean jump performance did not significantly improve following acute WBV of varying frequencies. This was true for CMJ height, peak force during CMJ take off phase, RFD and RSI. As discussed (see section 5.4.2), one possible reason could be the influence of BM on the WBV parameters of the stimulus provided from the platform. Participants from chapter 5 had a BM range of 73 to 101 kg therefore, it is hypothetically possible that the additional load of the participant standing on the platform may affect the stimulus delivered by the platform. Much of the previous WBV research did not consider this as a contributing factor and therefore did not investigate this aspect further.

Previous literature has utilised loaded WBV protocols involving participants BM (mean: 78 ± 2.1 kg) and up to 60 kg additional load (Ronnestad, 2009). As well as a participant mean BM of 110 ± 24 kg with additional loads of 100 kg (Ronnestad et al., 2012). Both studies investigated jump performance and EMG responses but neither of these studies investigated the possible effect of such total external loads (BM + additional load) on the WBV stimulus. BM is likely to affect WBV parameters as larger masses may dampen the magnitude of vibration stimulus (acceleration) generated by platforms (Pel *et al.*, 2009; Marin & Rhea, 2010b). At present the influence of these loads on WBV output is unknown and identifying limitations in WBV platform output is problematic at present due to a lack of research analysing the performance of WBV platforms (Preatoni et al., 2012).

In theory, frequency and D_{PTP} could be susceptible to the influences of additional load placed on the platform. It is these parameters which contribute to the acceleration or g-force exposed to WBV users (Wilcock *et al.*, 2009; Preatoni *et al.*, 2012). As such, any external interaction that can potentially change these parameters may influence the magnitude of acceleration and hence the WBV effect on EMG activity and jump performance responses (Crewther et al., 2004). A better understanding of these parameters is required to allow optimal WBV protocols to be developed (Pollock et al., 2010). When prescribing WBV as a training stimulus the parameters which contribute towards the magnitude of exposure should be carefully considered. As both frequency and D_{PTP} combine to calculate acceleration, it may be plausible that by altering one or both that the overall acceleration magnitude may change, but also the body's response to that change in WBV stimulus (Cook et al., 2011).

There are few studies which have attempted to investigate the body's response to altered parameters (e.g. D_{PTP}). Varying D_{PTP} did have a beneficial effect on CMJ performance (1 – 1.6 % increase in CMJ peak power, although there was a significant frequency interaction found) (Adams et al., 2009). However, the research did not involve investigating an interaction of an external source on D_{PTP}; but directly changing D_{PTP} at the WBV platform, as well as varying WBV frequencies (30 – 50 Hz). In addition, the work quoted a D_{PTP} range which was taken from the platform. Contradicting this research, Bazett-Jones *et al.* (2008b) reported no effect of altering D_{PTP} on CMJ performance. A strength of the research was the use of accelerometers to directly measure the acceleration output, although the locations of these were not clearly described. A further limitation of the work by Bazett-Jones *et al.* (2008b) was the fact that frequency was simultaneously changed (30 – 50 Hz), therefore it is unknown which parameter may have influenced CMJ performance.

A method advocated by Rauch *et al.* (2010) allows the direct measurement of the true output of a WBV platform not only in acceleration but also frequency parameters. This approach allows for the comparison of acceleration magnitudes from the platform (i.e. theoretical acceleration) to the actual magnitude exposed to participants (i.e. true acceleration). The protocol by Rauch *et al.* (2010) also allows the influence of BM on acceleration to be examined.

One such study (Crewther et al., 2004) utilised accelerometers to monitor the influence of changing D_{PTP} during different WBV frequencies on the acceleration magnitude output. Only, at higher frequencies (30 Hz) did altering D_{PTP} influence acceleration magnitude output (9.8 % increase in g forces at 30 Hz compared to 3.9 and 0.7 % increases at 20 and 10 Hz respectively). The use of a Galileo type WBV platform limits this research to 10 – 30 Hz, lower than most of the existing WBV research. Additional limitations include that changes to D_{PTP} were not achieved by external factors but by changing foot position to alter D_{PTP}, during which non-standardised footwear was worn, introducing a potential for dampening variability. In fact, the external load applied during varying D_{PTP} values remained constant (mean participant BM of 69.6 ± 12.2 kg). It could be argued that this represents a lower BM than "common" participants, especially in the trained e.g. chapter 5 participants.

Cook *et al.* (2011) utilised heavier participants (86 ± 9 kg) and reported a significant effect of varying D_{PTP} on acceleration. However, again no direct investigation was completed into the influence of external load on acceleration output. Further limiting the work was that 50 Hz was not utilised and the low knee flexion angle of 30 – 40 º may not represent commonly adopted postures. EMG recordings were not taken and therefore any neuromuscular effect of varying D_{PTP} on acceleration is unknown, as is the posture adopted.

A second study to recruit heavier participants (55 and 90 kg) completed a validation of acceleration output (Preatoni et al., 2012). It was reported that frequency was not external load dependent and the true frequency matched the theoretical frequency (< 2 Hz frequency average error). However, D_{PTP} and acceleration output were BM dependent with varying acceleration average error of 5 – 25 %. Limiting the findings were the recruitment of female only participants as there may be a gender specific response to WBV (Bazett-Jones *et al.*, 2008b; Marin & Rhea, 2010b). The validation protocol only involved two participants and 90 kg as the greatest BM; during which no EMG measures were taken. It could be argued that often WBV users may well have a greater mass than 90 kg (e.g. mean mass: 104.2 ± 13.2 kg, see methodology section 4.2.1). In addition, large variability in WBV stimulus received between participants of differing BM are likely (Marin & Rhea, 2010b) as highlighted in the large inter-individual differences reported in chapters 4 and 5.

One study which directly investigated the influence of external loading on acceleration, incorporated a wider range of WBV frequencies and higher external loads of 63, 82 and 100 kg (Pel et al., 2009). It was reported that external load had little influence as both theoretical and true accelerations matched closely (within a 6 – 10 % difference). However, a number of limitations exist such as unknown frequency and D_{PTP} parameters as detailed methodology was lacking. The posture adopted of 30 º knee flexion and no EMG recordings also limit the findings.

Pollock *et al.* (2010) recorded EMG measures and reported that changes in D_{PTP} influenced EMG activity, especially in the gastrocnemius and tibialis anterior muscles. However, no direct investigations were made into the influence of changing external load on D_{PTP}. In addition, the effect on vastus lateralis EMG recordings is unknown and the use of a pivotal type WBV platform limits the findings to 5 – 30 Hz frequencies only. This was highlighted as the low frequencies used elicited similar EMG activity as fast walking, suggesting a low magnitude of training stimulus (Pollock et al., 2010). The posture adopted of 15 º knee flexion is likely to have influenced the EMG activity, especially in quadriceps groups, had vastus lateralis EMG activity been recorded. Finally, the participants' training history was unclear.

Ritzmann *et al.* (2012) directly investigated the influence of increasing external load by means of a weighted bar. Increased EMG activity of soleus (+ 11 %), gastrocnemius (+ 14 %), rectus femoris (+ 21 %) and tibialis anterior (+ 13 %) muscles was found due to additional external load. This research investigated both a pivotal type (Galileo) and a tri-planar type (Powerplate) platform. However, to allow direct comparison of the two types of platform only 5 – 30 Hz frequencies were selected, limiting the findings. Again, vastus lateralis EMG activity was not recorded. More importantly, no direct measure of acceleration was utilised. Therefore, the influence of additional external load on acceleration output by the platforms could not be investigated simultaneously. Participants were also unlikely to be well-trained, described as “physically fit students” (Ritzmann *et al.*, 2012).

Previous research (Pel *et al.*, 2009; Pollock *et al.*, 2010) has also considered the transmission of acceleration output from WBV platforms through body segments. Pel *et al.* (2009) and Pollock *et al.* (2010) reported significant (84 – 97 %) reductions in acceleration magnitude at higher body segments (knee and hip). This suggests a dampening mechanism as the WBV stimuli travel through musculoskeletal structures. However these findings were based on WBV stimuli delivered by tri-planar and pivotal platforms.

Further research is required to investigate the influence of external loading on both EMG activity and acceleration output; recorded simultaneously during WBV frequencies commonly used in vertical WBV platform types (e.g. 30 – 50 Hz). Further investigation is required quantifying a possible dampening of WBV stimuli at higher body segments during vertical type WBV. This research is required in well-trained participants adopting larger knee flexion angle positions to elicit potential WBV responses under a degree of stretch in quadriceps muscle groups. Therefore, the objective of this study was to determine the influence of external loading on EMG response and acceleration during acute WBV at different frequencies; with an additional objective of determining whether a dampening effect occurs at higher body segments.

7.2 METHOD

7.2.1 Experimental approach

To investigate the influence of external loading on acceleration and EMG response during different WBV frequencies, a randomised single blinded study design was utilised. In this study, which eliminates potential order effects, frequency, external load and accelerometer location are the independent variables and acceleration and EMG response are the dependent variables.

Participants attended one test session, during which frequency and external load were altered in a randomised order for a total of 12 conditions. The location of three acceleration recording sites (platform, tibia and iliac) remained constant throughout. For 12 conditions participants experienced 20s of WBV (set at 3 mm D_{PTP}) at varying frequencies (0, 30, 40 and 50 Hz) over three external loads (80, 90 and 100 kg). For each condition both acceleration and EMG response were measured during which participants adopted a 90 º knee flexion squat, with 120 s recovery between.

7.2.2 Participants

A total of 10 trained male participants (age, 26 ± 5 years; height, 176 ± 6 cm; mass 73.4 ± 3.1 kg) volunteered to take part. Trained was defined as a minimum of six months structured strength and conditioning training. Additional inclusion and exclusion criteria have been previously outlined (see section 3.1); however a specific inclusion criterion to the study in chapter 7 was a BM of approximately 70 to 79 kg. This was to ensure participants' BM could be increase to the three external loads of 80, 90 and 100 kg.

7.2.3 Procedure

7.2.3.1 Warm up

A standardised warm up was utilised including 5 minutes cycling at a self-selected speed with 0.5 kg load (Monark 874E, Monark Ergometer, Sweden). Participants completed 10 repetitions of both slow and fast BM squats. This was modified from Ebben & Petushek (2010) and due to the nature of the protocol involved, only squatting activities were completed for specificity reasons and to maximise EMG response (Hough et al., 2009).

7.2.3.2 Electromyography

Right vastus lateralis, rectus femoris and gastrocnemius muscles were recorded. Electrode placement sites for these muscles are explained in general methods, (section 3.2). In addition, the following procedure was completed for this study only. Electrode placement remained constant throughout and EMG data was recorded from 5 to 15 s during 20 s WBV, as EMG signal has been demonstrated stable during sampling (see section 5.2.5.1). Excluding the first and last 2.5 s of WBV exposure prevented start-up and deceleration stages from being recorded and ensured a steady WBV output.

7.2.3.3 Accelerometers

A total of three accelerometers were utilised (see equipment section 7.2.4.2). Each accelerometer was calibrated by applying 1g (earth's gravity = 9.81 $m.s^{-2}$). As each accelerometer was tri-axial each accelerometer was calibrated to *z*-axis relative to global position (Pel et al., 2009). Each position was recorded using a standardised PVC box on which each accelerometer was securely attached by double sided tape. Each calibration position was standardised to level by a two axis spirit level and a 10 s sample was recorded. After these calibration positions all measurements relative to global position in the *z*-axis represented acceleration in the vertical direction, *y*-axis representing acceleration in the medial-lateral direction, and *x*-axis representing anterior-posterior direction (Cook et al., 2011).

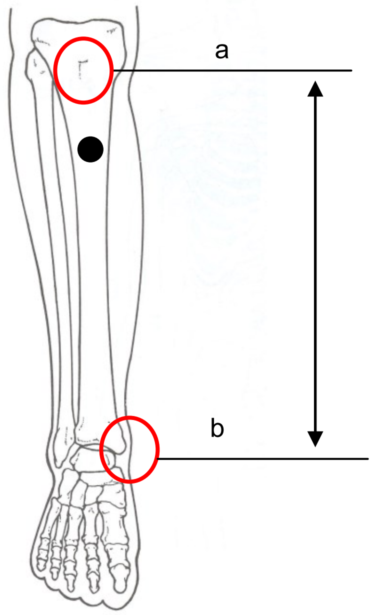
One accelerometer was placed in a standardised centre of the WBV platform, attached by double sided tape (Figure 7.1). As the WBV platform is portable a level position was standardised before each data collection. A second accelerometer was placed on the right anterior border of tibia, distal of tibial tuberosity, at a distance of 15 % of tibial tuberosity to medial malleolus (Thompson & Floyd, 1998), (Figure 7.1). Skin at this site was shaven to ensure secure fixation of accelerometer and prevent excessive non WBV-related movement (Crewther et al., 2004). Vertical placement was ensured by means of spirit level. Finally, a third accelerometer was placed on the right iliac crest (Figure 7.1). Clothing was moved and taped to ensure good contact for accelerometer fixation and that clothing did not make contact with the accelerometer. Placement of the tibial and iliac crest accelerometers aimed to record the acceleration at different body segments whilst being placed on bony anatomical landmarks to avoid soft tissue masses and potential error (Matsumoto & Griffin, 1998; Pel *et al.*, 2009).

Accelerometer orientation perpendicular to *z*-axis in 90 º knee flexion squat posture was ensured by means of spirit level. Accelerometer placement was maintained throughout all the 12 conditions, and recorded throughout the session.

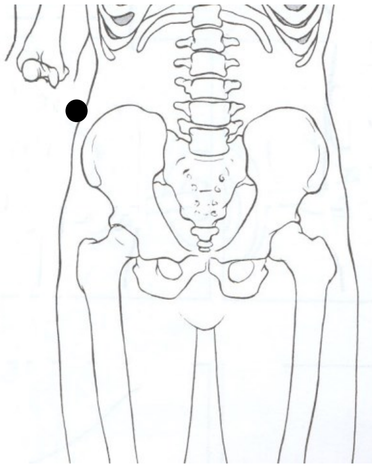
Figure 7.1: Accelerometer location sites, anatomical diagrams taken from Thompson & Floyd (1998).

Standardised location of accelerometer placed in the centre of the platform.

Tibia accelerometer location 15% of the distance between tibial tuberosity (a) and medial malleolus (b), secured onto anterior border of tibia.



Iliac crest accelerometer location.



7.2.3.4 External load

To achieve three standardised external loads each participant was measured without shoes to determine actual BM (Seca, model 780, Hamburg, Germany). The additional weight required to obtain 80, 90 and 100 kg standardised external loads was calculated. The external loads were selected to represent both the BM range of those participants recruited in chapter 5; but also "typical" WBV users. Participants wore a weighted vest to reach 80 kg external load and then attached 10 kg weight(s) to reach 90 and 100 kg external loads during WBV exposure. During recovery all additional mass was removed.

7.2.3.5 Whole body vibration

Please refer to general method section 3.3 for details of WBV procedures. The specific WBV protocol unique to this present study consisted of 20 s WBV exposure at either 0, 30, 40 and 50 Hz, depending on the condition (0 Hz acting as control). Participants were blinded to which frequency they were exposed to. Participants adopted 90 º knee flexion squat (Cardinale & Lim, 2003b). Foot placement in relation to the WBV platform was shoulder width apart equidistance from the platform centre, and was standardised for each condition by marking the location of lateral border of each foot.

7.2.4 Equipment

7.2.4.1 Electromyography

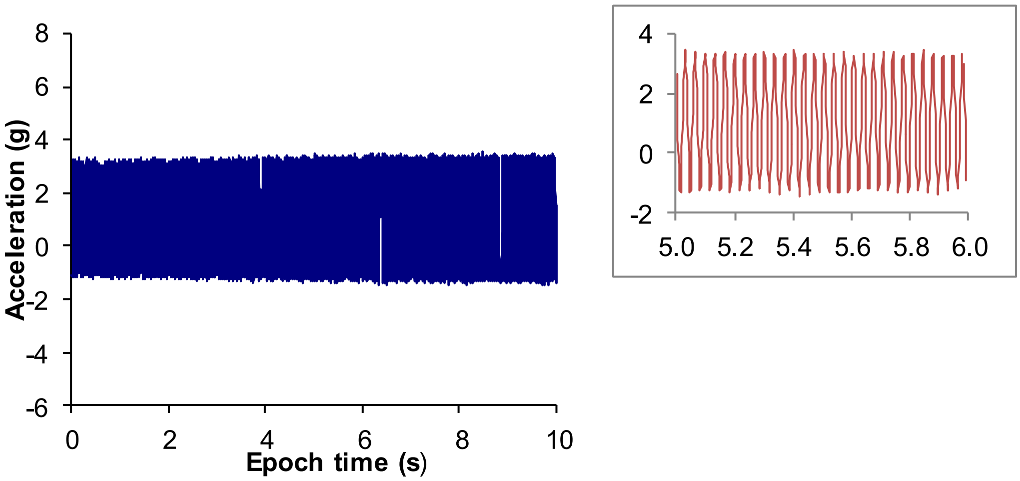
For detailed description of EMG equipment utilised in the current study, please refer to general method section 3.2.2.

7.2.4.2 Accelerometers

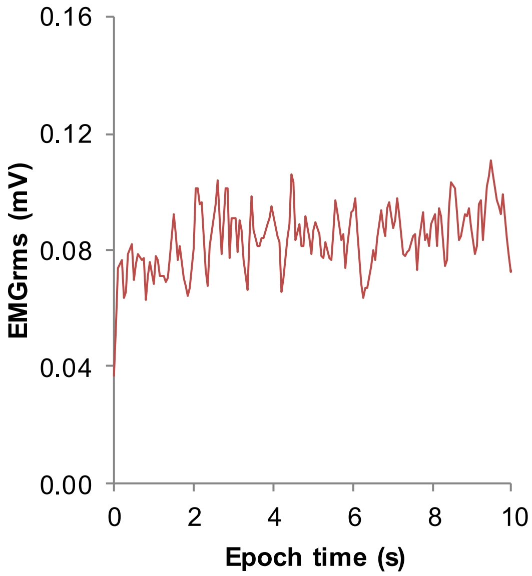
Acceleration data was recorded via tri-axial accelerometers (USB Impact Accelerometer, model X250-2, Gulf Coast Data Concepts, USA). The X250-2 model is approximately 10 cm x 2 cm x 2 cm (length x width x height) and weighs 33 grams. Acceleration data was recorded at a sampling rate of 512 Hz onto a 1 GB microSD card at high gain and 14-bit resolution. Figure 7.2 illustrates a representative participant's platform acceleration and vastus lateralis EMG_{rms} recorded during 30 Hz WBV and 80 kg external load.

Figure 7.2: Representative participant's platform acceleration and vastus lateralis EMG_{rms} (mV) over a measured epoch during 30 Hz WBV with 80 kg external load.

A. Platform acceleration data



B. EMG_{rms} data



7.2.4.3 Weighted vest

To standardise external loads, a weight vest (Alex Athletics, UK) was utilised. An empty vest weighed 0.5 kg and contained nine compartments in which 1.1 kg weights could be inserted. Therefore, from participant's actual BM the first standardised external load (80 kg) was achieved to the nearest 0.1 kg. To achieve 90 kg External load one Olympic disc 10 kg was attached to the weight vest via secure ropes and karabiners. To achieve 100 kg External load, a second Olympic disc 10 kg was attached. Olympic weights were attached on the posterior portion of the vest so hung in mid-air during squatting.

7.2.4.4 Whole body vibration

As the WBV was delivered by the same platform as past studies, details are covered in the general methods section, (see section 3.3).

7.2.5 Data analysis

7.2.5.1 Electromyography

In addition to EMG data analysis as described in general methods, the following study specific data analysis was completed. A 14th Order Butterworth band pass filter (20-300 Hz) and a 14th Order Butterworth notch filter (band stop filter ± 1.5 Hz centred at WBV fundamental frequency and relevant sub harmonics) were applied to raw EMG data. This was based on the findings of chapter 6, section 6.3 in which this type of notch filter technique effectively removed motion artifacts without excluding true muscle signal. Data was root mean squared (EMG_{rms}) and transferred to excel spreadsheets for further analysis which included determining mean EMG_{rms} values during 10 s WBV for each of the 12 conditions. As EMG data was collected during a single session in a repeated measures study design there was no requirement for normalisation (Enoka, 2002).

7.2.5.2 Acceleration data

Acceleration data recorded during the 12 x 10s conditions were identified (80, 90 and 100 kg external loads x 0, 30, 40 and 50 Hz WBV frequencies). Acceleration data from three conditions involving 0 Hz will not be presented as acceleration magnitude is taken as 9.81 $m.s^{-2}$ (1 g, the earth's gravitational field). For the remaining nine conditions, the first and last 5s were excluded from subsequent data analysis. Acceleration data from *x*, *y* and *z* – axis were imported into excel in which the raw data count was converted to g units using the following equation;

$$(\text{counts} - 8192) \times \frac{70}{16384}$$

(Gulf Coast Data Concepts, 2009)

Data was converted from g units into acceleration, ($m.s^{-2}$) by multiplying by earth's gravity (9.81 $m.s^{-2}$). Peak accelerations during 10s of WBV were identified via an Excel algorithm and then averaged for each 10s period.

Due to the tri-axial nature of acceleration data recording the resultant acceleration (A_R) was calculated using a Pythagoras Theorem approach to calculate resultant from three vectors (Kreighbaum & Barthels, 1996), (appendix 7).

To further investigate the influence of external loading on WBV accelerations unidirectional axis data was explored. *Z*-axis representing vertical acceleration (A*z*), *y*-axis representing medial-lateral acceleration (A*y*) and *x*-axis representing anterior-posterior acceleration (A*x*). A*z* represents the theoretical output of this type of WBV platform and accordingly, A*z* will be explored in detail, whereas A*y* and A*x* data will be briefly summarised and expressed as a percentage of A_R.

Using A_R, transmission ratios were calculated which included; iliac to platform ratio (A_R I : P); tibia to platform ratio (A_R T : P); and iliac to tibia ratio (A_R I : T). Thus allowing the dampening in leg segments during WBV to be quantified (Cook et al., 2011).

7.2.6 Statistical analysis

7.2.6.1 Electromyography

The effect of external load on EMG_{rms} of vastus lateralis, rectus femoris and gastrocnemius during different WBV frequencies was analysed by means of a 2-way repeated measures ANOVA (frequency [0, 30, 40, 50 Hz] x external load [80, 90, 100 kg]).

7.2.6.2 Acceleration

The effect of external load on A_R and Az at different accelerometer locations over different WBV frequencies was analysed by means of a 3-way repeated-measures ANOVA (frequency [30, 40, 50 Hz] x accelerometer location [platform, tibia, iliac] x external load [80, 90, 100 kg]). If a significant interaction effect was found a 2-way repeated-measure ANOVA was utilised, in which significant main effect *F*-values were further analysed by post hoc *t*-tests as described previously. If significant 2-way ANOVA interaction effects were found, a 1-way ANOVA followed by post hoc *t* – tests were utilised to further explore interactions.

The effect of external load on A_R I : P, A_R T : P and A_R I : T over different WBV frequencies was analysed by means of a 2-way repeated-measures ANOVA (frequency [30, 40, 50 Hz] x external load [80, 90, 100 kg]).

7.2.6.3 Relationship between acceleration and electromyography

A possible relationship between platform A_R and EMG responses of three muscles were explored via 2-tailed Pearson Correlations. To investigate A_R I : P and EMG correlations, vastus lateralis and rectus femoris EMG data were selected. To investigate A_R T : P and EMG correlations gastrocnemius EMG was data were selected.

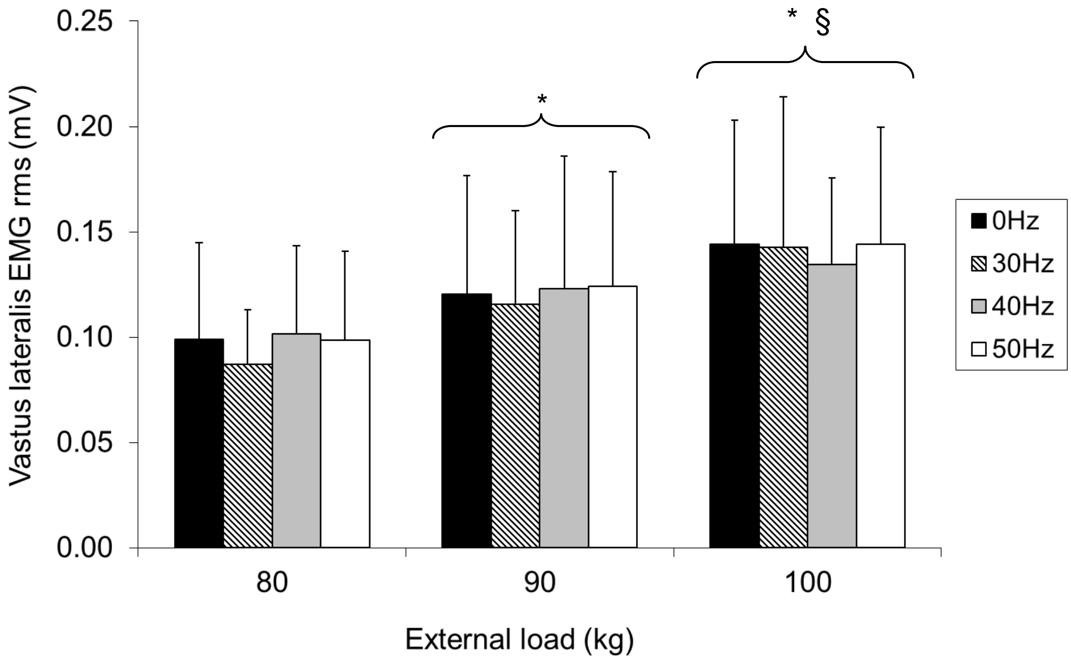
7.3 RESULTS

7.3.1 Electromyography data

7.3.1.1 Vastus lateralis

A significant main effect for external load was found (F [1.16,10.41] = 30.83, p = 0.001), with a very large effect size (r = 0.88), Figure 7.3. Post hoc analysis revealed significantly higher vastus lateralis EMG_{rms} between: 80 and 90 kg (mean difference 0.02 mV, $p < 0.05$); 80 and 100 kg (mean difference 0.05 mV, $p < 0.001$); 90 and 100 kg (mean difference 0.02 mV, $p < 0.001$). As expected vastus lateralis EMG_{rms} increased with increasing external loads. No significant main effect for frequency was found (F [1.57,14.11] = 1.14, p = 0.33), with a moderate effect size (r = 0.33). No significant frequency x external load interaction effect was found (F [1.17,10.54] = 0.53, p = 0.51), with a small effect size (r = 0.24).

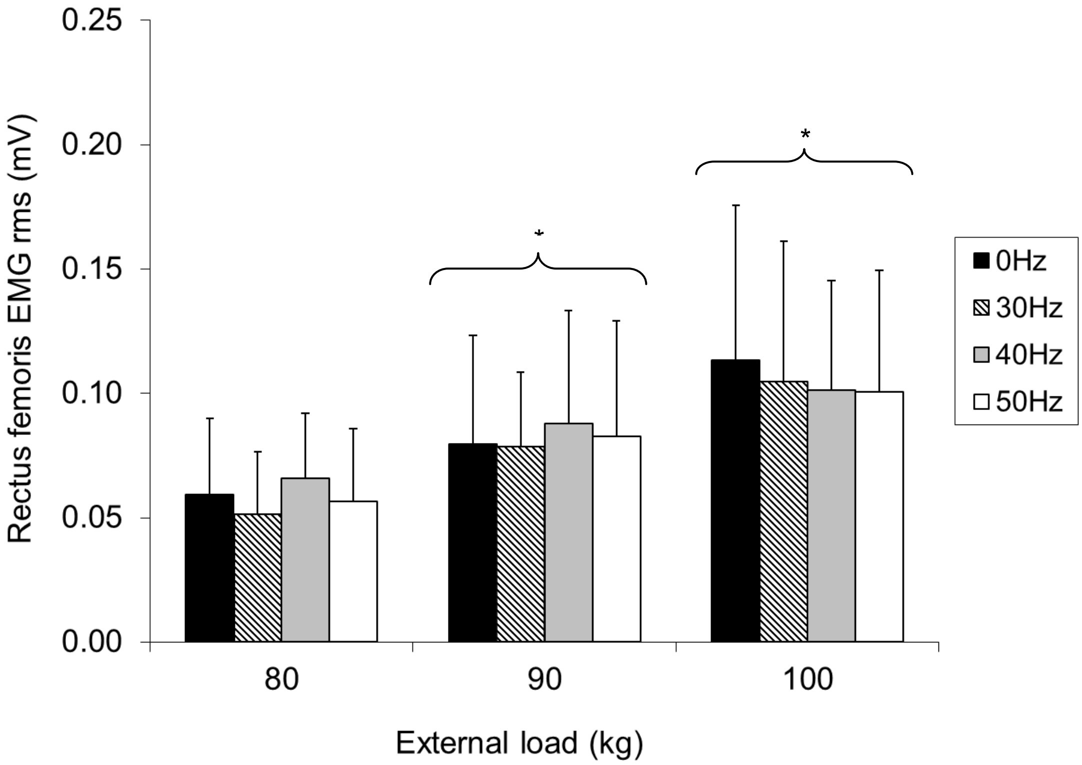
Figure 7.3: Mean ± SD vastus lateralis EMG_{rms} (mV) during four different WBV frequencies and three external loads. A significant main effect for external load was found. *, significantly higher than 80 kg; §, significantly higher than 90 kg. No main effect for frequency was found.



7.3.1.2 Rectus femoris

As with vastus lateralis there was a significant main effect for external load, similar to that of vastus lateralis (F [2,18] = 13.71, $p < 0.001$), with a very large effect size (r = 0.77), Figure 7.4. Post hoc analysis revealed significantly higher rectus femoris EMG_{rms} between: 80 to 90 kg (mean difference 0.02 mV, $p < 0.05$); 80 to 100 kg (mean difference 0.05 mV, $p < 0.01$). However, unlike vastus lateralis there was no significant difference between 90 and 100 kg ($p = 0.08$). There was no significant main effect for frequency found (F [1.85,16.65] = 1.49, $p = 0.25$), with a moderate effect size (r = 0.37). No significant frequency x external load interaction effect was found (F [1.79,16.10] = 0.96, $p = 0.39$), with a moderate effect size (r = 0.32).

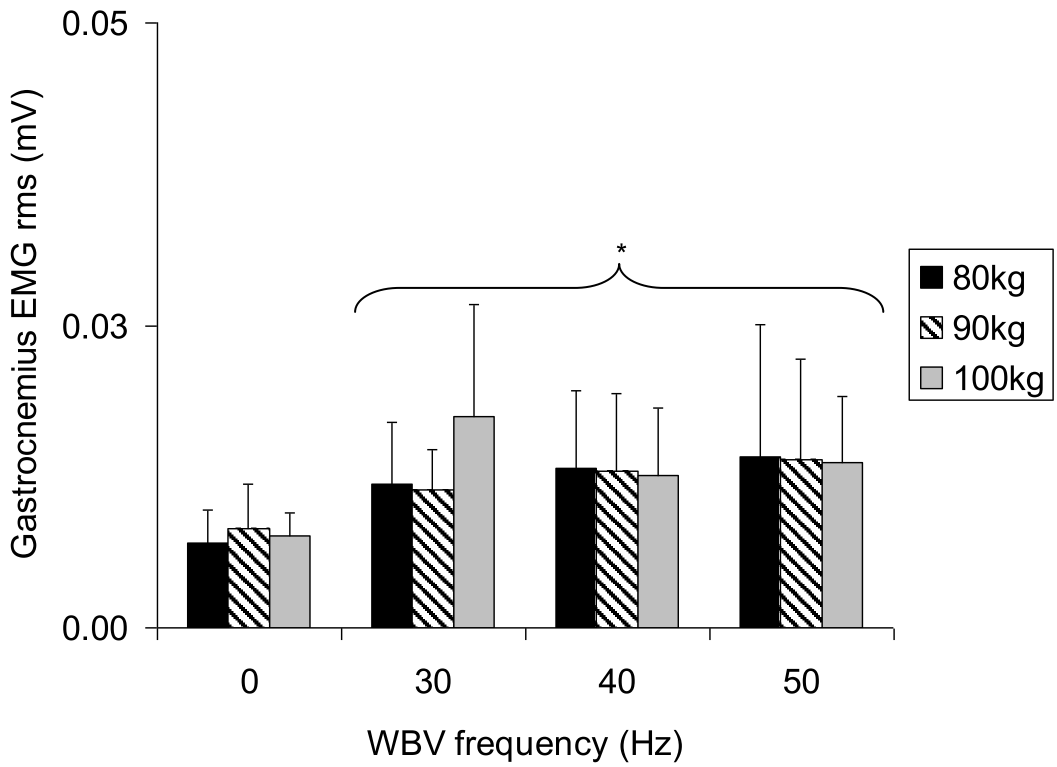
Figure 7.4: Mean ± SD rectus femoris EMG_{rms} (mV) during four different WBV frequencies and three external loads. A significant main effect for external load was found. *, significantly higher than 80 kg. No main effect for frequency was found.



7.3.1.3 Gastrocnemius

In contrast to vastus lateralis and rectus femoris, no significant main effect for external load was found (F [1.16,10.44] = 0.61, p = 0.48), with a small effect size r = 0.24). No significant frequency x external load interaction effect was found (F [1.64,14.73] = 1.09, p = 0.35), with a moderate effect size r = 0.33). Also in contrast to both vastus lateralis and rectus femoris, a significant main effect for frequency was found in gastrocnemius EMG_{rms} (F [3,27] = 9.06, p = 0.001), with a very large effect size (r = 0.71), Figure 7.5. Post hoc analysis revealed significant increased EMG_{rms} between: 0 and 30 Hz (mean differences, 0.01 mV, $p < 0.01$); 0 and 40 Hz (mean differences, 0.01 mV, $p < 0.05$); and 0 and 50 Hz (mean differences, 0.01 mV, $p < 0.05$). No significant differences between 30, 40 and 50 Hz WBV frequencies were found ($p > 0.05$).

Figure 7.5: Mean ± SD gastrocnemius EMG_{rms} (mV) during four different WBV frequencies and three external loads. A significant main effect for frequency was found. *, significantly higher than 0 Hz. For illustrative purposes the data is presented grouped by WBV frequency.

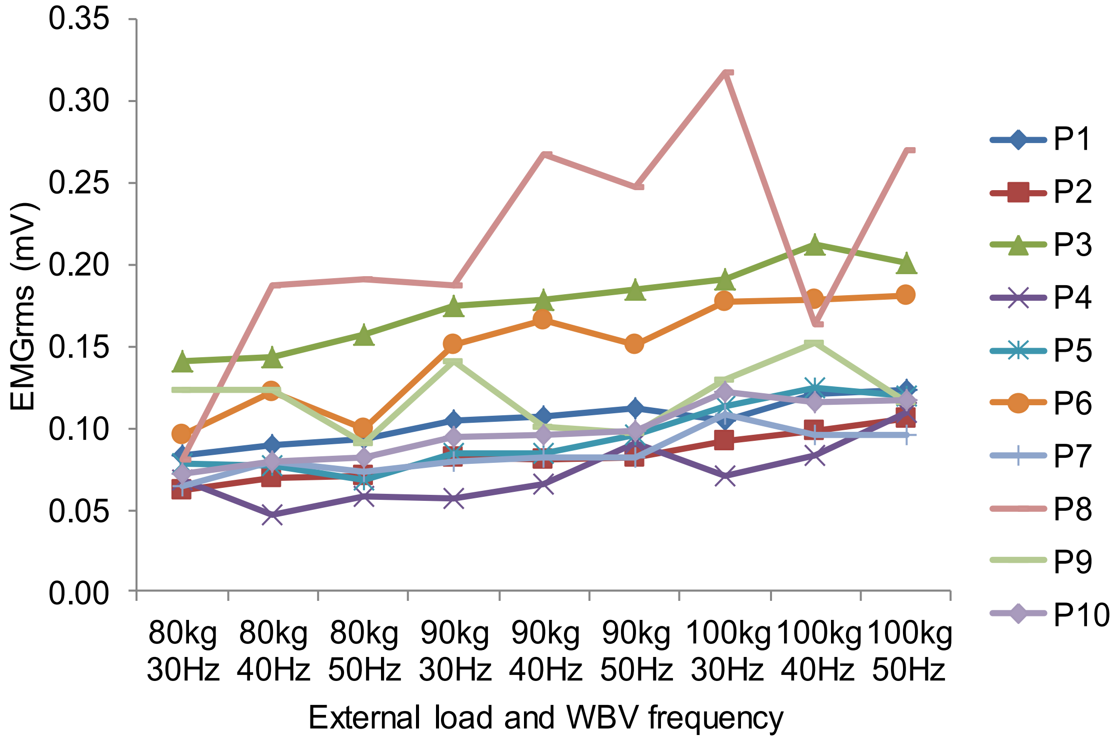


7.3.1.4 Individual responses in muscle activity

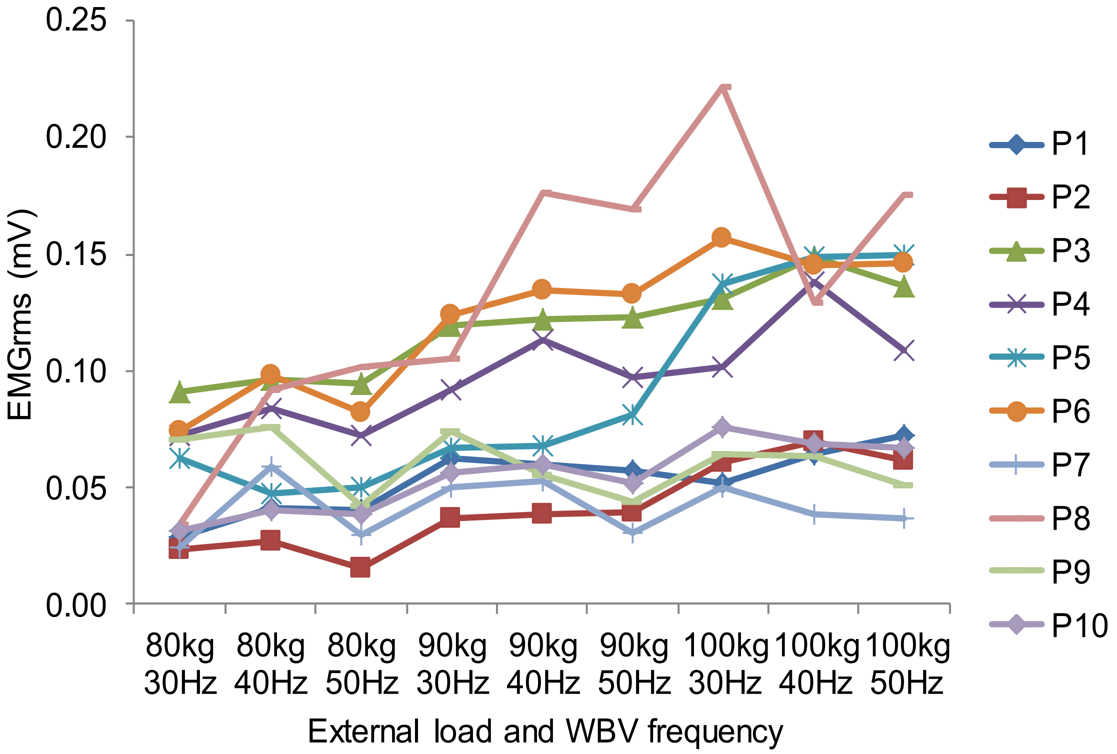
Each participant's EMG_{rms} response across WBV frequency and external load for vastus lateralis, rectus femoris and gastrocnemius muscle groups are presented in figure 7.6.

Figure 7.6: Individual EMG_{rms} responses during different WBV frequencies and external loads of: A, vastus lateralis; B, rectus femoris; C, gastrocnemius

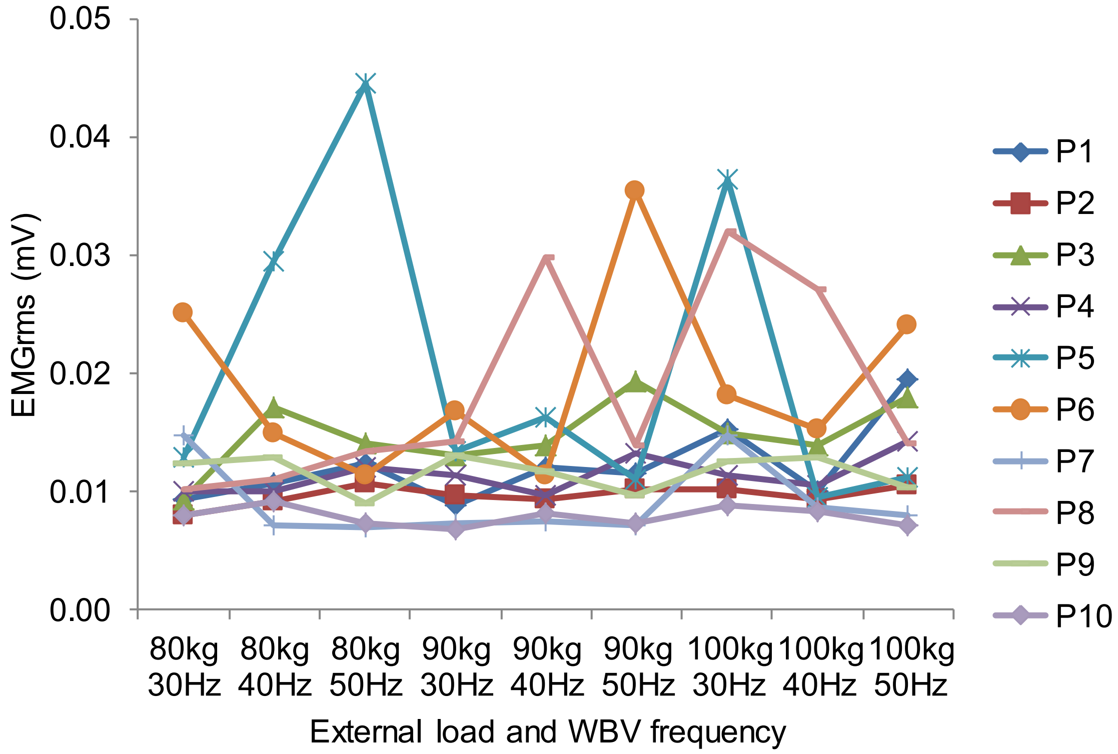
A



B



C



7.3.2 Acceleration data

7.3.2.1 Resultant acceleration

7.3.2.1.1 3-way ANOVA resultant acceleration (frequency x accelerometer location x external load)

In contrast to EMG data, no significant main effect was found for external load (F [2,18] = 2.98, p = 0.08), with a moderate effect size (r = 0.50).

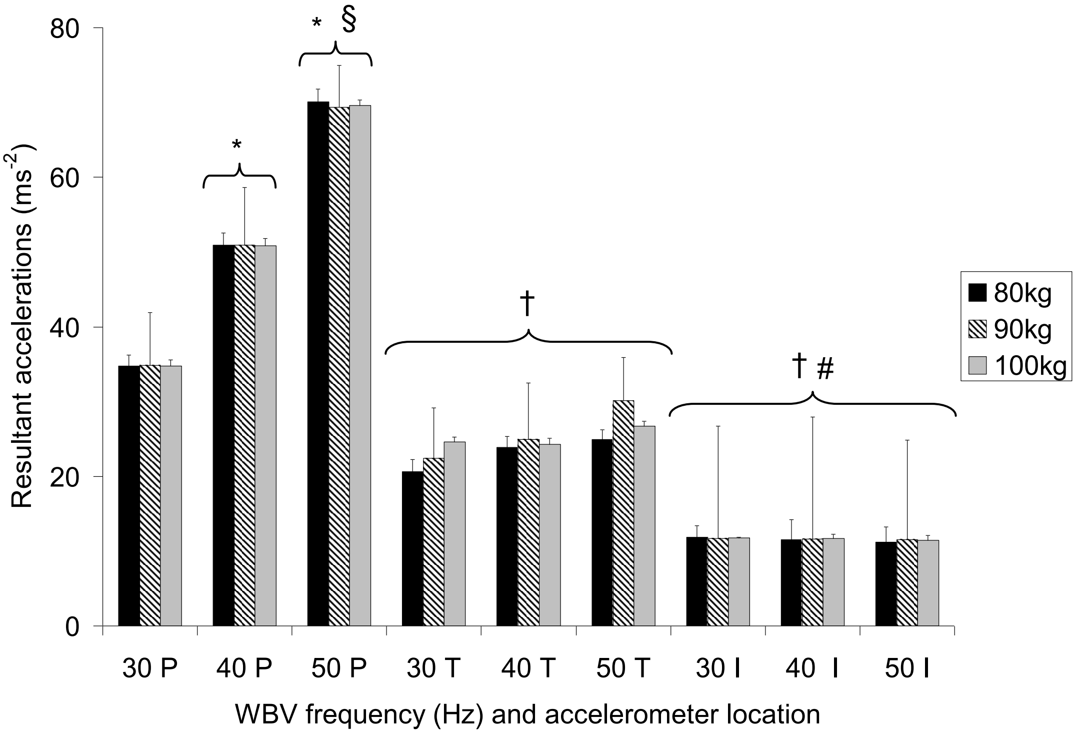
However, a significant main effect for frequency was found (F [1.06, 9.58] = 81.56, $p < 0.001$) with a near perfect effect size (r = 0.95), Figure 7.7. Post hoc analysis revealed significant increases in A_R during: 30 Hz compared to 40 Hz (mean difference 5.92 m.s^{-2}, $p < 0.001$); 30 Hz compared to 50 Hz (mean difference 13.07 m.s^{-2}, $p < 0.001$); and 40 Hz compared to 50 Hz (mean difference 7.15 m.s^{-2}, p <

0.001). These results suggest that, as expected, A_R increases at higher WBV frequencies.

A significant main effect was also found for accelerometer location (*F* [1.03, 9.23] = 205.86, $p < 0.001$) with a near perfect effect size ($r = 0.98$), Figure 7.7. Post hoc analysis revealed significant decreases in A_R between platform and tibia level (mean difference 27.08 $m.s^{-2}$, $p < 0.001$) and between platform and iliac level (mean difference 40.21 $m.s^{-2}$, $p < 0.001$). There was also a significant decrease in A_R recorded between tibia and iliac level (mean difference 13.13 $m.s^{-2}$, $p < 0.01$). These results suggest that at superior body segments, lower A_R was observed.

Significant frequency x accelerometer location interaction effect was found (*F* [1.07, 9.64] = 44.03, $p < 0.001$) with near perfect effect size ($r = 0.91$), indicating that the location of accelerometer had differing effects on A_R depending on what frequency of WBV was used. No interactions were significant when external load was included ($F = 0.83 – 3.38$, $p = 0.08 – 0.43$) with small to moderate effect sizes ($r = 0.28 – 0.52$).

Figure 7.7: Mean ± SD peak resultant acceleration (A_R) ($m.s^{-2}$) for platform, tibia and iliac crest accelerometer locations during three different WBV frequencies (30, 40 and 50 Hz) and external loads (80, 90 and 100 kg). Significant main effects for frequency (* and §) and accelerometer location († and #) and significant interaction effects for frequency x accelerometer location (all $p < 0.01$). P, platform; T, tibia; I, iliac; *, significantly higher than 30 Hz at the same location; §, significantly higher than 40 Hz; †, significantly lower than platform location; #, significantly lower than tibia location.



7.3.2.1.2 2-way ANOVA resultant acceleration (frequency x accelerometer location)

As anticipated, significant main effects for frequency and accelerometer location were found over all three external loads, Table 7.1. Significant frequency x accelerometer location interaction effects were found over all three external loads, (Table 7.1) and therefore were explored further via 1-way ANOVA (section 7.3.2.1.3).

Table 7.1: 2-way repeated measures ANOVA for A_R (frequency [30, 40, 50 Hz] x accelerometer location [platform, tibia, iliac]) applied at three external loads (80, 90, 100 kg). F, F value; df, degrees of freedom; sig. (*p*), significance p value; Acc location, accelerometer location.

External load (kg)	Variable	F (df)	sig. (*p*)	Effect size (*r* =)
80	Frequency	49.21 [1.22,10.96]	< 0.001	0.92
80	Acc location	219.44 [1.04,9.35]	< 0.001	0.98
80	Frequency x acc location	33.48 [1.26,11.31]	< 0.001	0.89
90	Frequency	50.20 [1.14,10.26]	< 0.001	0.92
90	Acc location	158.57 [1.03,9.27]	< 0.001	0.97
90	Frequency x acc location	25.67 [1.14,10.29]	< 0.001	0.86
100	Frequency	97.05 [1.11,10.03]	< 0.001	0.96
100	Acc location	212.25 [1.02,9.17]	< 0.001	0.98
100	Frequency x acc location	54.89 [1.08,9.72]	< 0.001	0.93

7.3.2.1.3 1-way ANOVA resultant acceleration (frequency x accelerometer location)

As frequency x accelerometer location interaction effects were significant over all external loads, 1-way ANOVA's were completed exploring first, the effect of frequency at three accelerometer locations and secondly; the effect of accelerometer location at three frequencies. These were completed for all external loads.

Only at platform level did higher WBV frequencies significantly increase A_R, ($p <$ 0.001 and near effect size $r = 1.00$). Significant increases were found at: platform level during 30 versus 40 Hz; 30 versus 50 Hz; and 40 versus 50 Hz. This was evident across all external loads as anticipated, Table 7.2. In contrast, at tibia level, higher frequencies did not significantly affect A_R at any of the three external loads, Table 7.2. At iliac level significant main effects were detected at 80 kg external load however, no post hoc differences were found, Table 7.2. At 90 and 100 kg external loads higher frequencies did not significantly affect A_R.

Table 7.2: 1-way ANOVA frequency x accelerometer location interactions, the effect of WBV frequency at three accelerometer locations. Difference in mean acceleration is presented as an increase relative to the lower frequency. ✓, significance level p < 0.001; (diff in mean acc), mean difference in acceleration ($m.s^{-2}$); X, not significant p > 0.05; ✓*, significance level p < 0.05 however, no post hoc differences found.

External load (kg)	WBV frequency (Hz)	Accelerometer location Platform (diff in mean acc)	Tibia	Iliac
80	30 vs. 40	✓ (16.16)	X	✓ *
	30 vs. 50	✓ (35.33)	X	✓ *
	40 vs. 50	✓ (19.17)	X	✓ *
90	30 vs. 40	✓ (16.11)	X	X
	30 vs. 50	✓ (34.54)	X	X
	40 vs. 50	✓ (18.43)	X	X
100	30 vs. 40	✓ (16.10)	X	X
	30 vs. 50	✓ (34.83)	X	X
	40 vs. 50	✓ (18.73)	X	X

In support, for all frequencies, A_R was significantly lower at tibia versus platform and iliac versus platform levels, (all p < 0.001) range of very large to near perfect effect sizes (r = 0.94 – 0.98), Table 7.3. This was evident at all three external loads. At all frequencies, A_R was significantly lower at iliac versus tibia level, Table 7.3. This was evident at all three external loads, except at 50 Hz WBV frequency during 80 kg external load.

Table 7.3: 1-way ANOVA frequency x accelerometer location interactions, the effect of accelerometer location at three WBV frequencies. Difference in mean acceleration is presented as a decrease relative to the lower frequency. P, platform; T, tibia; I, iliac; ✓, significance level (30 Hz p < 0.05, 40 Hz p < 0.001); X, not significant (p > 0.05); (mean diff in acc), mean difference in acceleration ($m.s^{-2}$).

External load (kg)	Accelerometer location	WBV frequency (Hz) 30 (mean diff in acc)	40 (mean diff in acc)	50 (mean diff in acc)
80	P vs. T	✓ (14.15)	✓ (27.10)	✓ (45.20)
	P vs. I	✓ (22.97)	✓ (39.44)	✓ (58.95)
	T vs. I	✓ (8.82)	✓ (12.34)	X
90	P vs. T	✓ (12.40)	✓ (26.02)	✓ (39.20)
	P vs. I	✓ (23.12)	✓ (39.30)	✓ (57.85)
	T vs. I	✓ (10.72)	✓ (13.28)	✓ (18.64)
100	P vs. T	✓ (10.15)	✓ (26.61)	✓ (42.89)
	P vs. I	✓ (22.96)	✓ (39.14)	✓ (58.14)
	T vs. I	✓ (12.81)	✓ (12.53)	✓ (15.25)

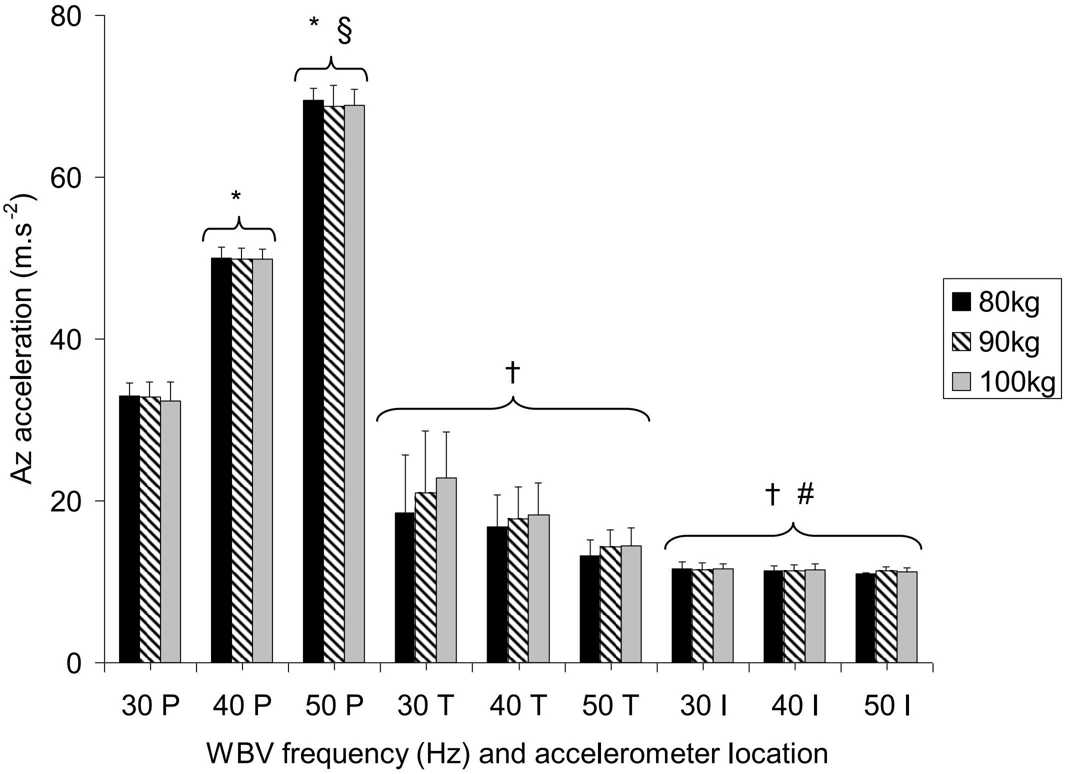
7.3.2.2 Vertical acceleration

7.3.2.2.1 3-way ANOVA vertical acceleration (frequency x accelerometer location x external load)

The results for 3-way ANOVA main effects and interaction effects on A*z* (see Figure 7.8) largely replicated the pattern of A_R. However, in contrast to A_R:

- A significant main effect for external load was found (F [2,18] = 3.70, $p < 0.05$) with large effect size ($r = 0.53$). However, no post hoc differences were found.
- A significant accelerometer location x external load interaction effect was found (F [4,36] = 8.97, $p < 0.001$) with very large effect size ($r = 0.71$), indicating that external load had differing effects on A*z* depending on the location of accelerometer and were further explored (section 7.3.2.2.4).

Figure 7.8: Mean ± SD peak A*z* acceleration ($m.s^{-2}$) for platform, tibia and iliac crest accelerometer locations during three different WBV frequencies (30, 40 and 50 Hz) and external loads (80, 90 and 100 kg). Significant main effects for frequency (* and §) and accelerometer location († and #) were found (p < 0.001). Significant interaction effects for frequency x accelerometer location and accelerometer location x external load were found (p < 0.001). P, platform; T, tibia; I, iliac; *, significantly higher than 30 Hz at the same location; §, significantly higher than 40 Hz; †, significantly lower than platform location; #, significantly lower than tibia location.



7.3.2.2.2 2-way ANOVA vertical acceleration (frequency x accelerometer location)

The results for 2-way ANOVA main effects and interaction effects of frequency x accelerometer location on A*z* exactly replicated the pattern of A_R; and therefore were explored further via 1-way ANOVA.

7.3.2.2.3 1-way ANOVA vertical acceleration (frequency x accelerometer location)

The results for 1-way ANOVA of frequency x accelerometer location on A*z* (see Table 7.4) largely replicated the pattern of A_R, with the following exception:

- At tibia level, higher frequencies significantly increased A*z* ($p < 0.05$ with large to very large effect sizes $r = 0.55 – 0.88$ across all three external loads). Post hoc analysis revealed at 80 and 90 kg external loads only 40 versus 50 Hz differences were significant, but at 100 kg external load significant increases existed at 30 versus 40 Hz, 30 versus 50 Hz and 40 versus 50 Hz, Table 7.4.

Table 7.4: 1-way ANOVA for A*z* frequency x accelerometer location interactions, the effect of WBV frequency at three accelerometer locations. Difference in mean acceleration is presented as an increase relative to the lower frequency. ✓, significance level $p < 0.001$; (diff in mean acc), mean difference in acceleration ($m.s^{-2}$); X, not significant $p > 0.05$; ✓*, significance level $p < 0.05$; ✓#, significance level $p < 0.05$ however, no post hoc differences found.

External load (kg)	WBV frequency (Hz)	Accelerometer location		
		Platform (diff in mean acc)	Tibia	Iliac
80	30 vs. 40	✓ (16.98)	X	✓ #
	30 vs. 50	✓ (36.52)	X	✓ #
	40 vs. 50	✓ (19.54)	✓ * (3.55)	✓ #
90	30 vs. 40	✓ (17.12)	X	X
	30 vs. 50	✓ (35.93)	X	X
	40 vs. 50	✓ (18.81)	✓ * (3.45)	X
100	30 vs. 40	✓ (17.43)	✓ (4.64)	X
	30 vs. 50	✓ (36.53)	✓ (8.45)	X
	40 vs. 50	✓ (19.10)	✓ (3.81)	X

7.3.2.2.4 2-way ANOVA vertical acceleration (accelerometer location x body mass load)

In contrast to A_R significant accelerometer location x external load interaction effects were detected and therefore were explored further by means of 2-way ANOVA.

No significant accelerometer location main effect was found at any WBV frequency, Table 7.5. Significant external load main effects were found at all three WBV frequencies, Table 7.5. Significant accelerometer location x external load interaction effects were found at 30 and 50 Hz WBV frequencies but not 40 Hz, see Table 7.5. Indicating that, unlike A_R, external load had differing effects on A*z* depending on the accelerometer location during 30 and 50 Hz only. Therefore, these were explored further via 1-way ANOVA, (see below).

Table 7.5: 2-way repeated measures ANOVA for Az (accelerometer location [platform, tibia, iliac] x external load [80, 90, 100 kg]) applied at three WBV frequencies (30, 40 and 50 Hz). F, F value; df, degrees of freedom; sig. (*p*), significance p value; acc location, accelerometer location; load, external load.

WBV frequency (Hz)	Variable	F [df]	sig. (*p*)	Effect size (*r* =)
30	Acc location	1.68 [2,18]	= 0.22	0.16
30	Load	76.93 [1.08,9.71]	< 0.001	0.90
30	Acc location x load	4.12 [4,36]	< 0.01	0.31
40	Acc location	1.22 [2,18]	= 0.32	0.12
40	Load	815.58 [1.11,10.01]	< 0.001	0.99
40	Acc location x load	1.45 [1.98,17.78]	= 0.26	0.14
50	Acc location	0.82 [2,18]	= 0.46	0.08
50	Load	3797.25 [2,18]	< 0.001	0.99
50	Acc location x load	4.23 [4,36]	< 0.01	0.32

7.3.2.2.5 1-way ANOVA vertical acceleration (accelerometer location x external load)

A series of 1-way ANOVA's first analysed the effect of accelerometer location at three external loads and second, the effect of external load at three accelerometer locations. These were only completed for 30 and 50 Hz WBV frequencies. These findings are in contrast to A_R as no interaction effects involving external loads were found at 3-way ANOVA level.

At all three external loads, significantly lower A*z* were recorded at higher body segment accelerometer location (all $p < 0.001$) and range of near perfect effect sizes ($r = 0.92 - 1.00$). Significant decreases in A*z* were found at 80, 90 and 100 kg external loads between platform versus tibia, platform versus iliac and tibia versus iliac locations. This was evident only for 30 and 50 Hz, Table 7.6. However, for all accelerometer locations, there was no significant change in A*z* with increasing external loads, Table 7.7.

Table 7.6: 1-way ANOVA for Az accelerometer location x external load interactions, the effect of accelerometer location at three external loads. Difference in mean acceleration is presented as a decrease relative to the first accelerometer location. P, platform; T, tibia; Iliac, I; ✓, significance level (p < 0.001); (mean diff in acc), mean difference in acceleration ($m.s^{-2}$).

WBV frequency	Accelerometer location	External load (kg) 80 (mean diff in acc)	90 (mean diff in acc)	100 (mean diff in acc)
30	P vs. T	✓ (14.51)	✓ (11.81)	✓ (9.53)
	P vs. I	✓ (21.37)	✓ (21.30)	✓ (20.86)
	T vs. I	✓ (6.87)	✓ (9.50)	✓ (11.33)
50	P vs. T	✓ (56.32)	✓ (54.44)	✓ (54.51)
	P vs. I	✓ (58.50)	✓ (57.40)	✓ (57.71)
	T vs. I	✓ (2.18)	✓ (2.96)	✓ (3.20)

Table 7.7: 1-way ANOVA for Az accelerometer location x external load interactions, the effect of external load at three accelerometer locations. Difference in mean acceleration is presented as a decrease relative to the first accelerometer location. P, platform; T, tibia; Iliac, I; X, not significant p > 0.05; ✓*, significance level p < 0.05 however, no post hoc differences found; (mean diff in acc), mean difference in acceleration ($m.s^{-2}$).

WBV frequency	External load (kg)	Accelerometer location P (mean diff in acc)	T (mean diff in acc)	I (mean diff in acc)
30	80 vs. 90	X	X	X
	80 vs. 100	X	X	X
	90 vs100	X	X	X
50	80 vs. 90	✓ *	✓ *	✓ *
	80 vs. 100	X	X	X
	90 vs100	X	X	X

7.3.2.3 Platform medial-lateral and anterior-posterior accelerations

For reference, platform A*y* and A*x* should theoretically be 0 $m.s^{-2}$ for a vertical oscillating platform. Mean peak platform A*y* and A*x* during three WBV frequencies are presented in Table 7.8. As a percentage of A_R, A*y* ranged from 12.4 to 32.7 %

characterised by an inverse relationship with frequency, Table 7.8. A*x* ranged from 3.8 to 10 % of A_R, again inversely with WBV frequency, Table 7.8.

Table 7.8: Platform medial-lateral (A*y*) and anterior-posterior (A*x*) accelerations expressed in magnitude ($m.s^{-2}$) and as a percentage of A_R during three frequencies of WBV.

WBV frequency (Hz)	External load (kg)	A*y* ($m.s^{-2}$)	A*y* (% A_R)	A*x* ($m.s^{-2}$)	A*x* (% A_R)
30	80	9.91	28.5	3.24	9.3
	90	10.56	30.3	3.40	9.8
	100	11.36	32.7	3.47	10.0
40	80	9.27	18.2	3.22	6.3
	90	9.27	18.2	3.45	6.8
	100	9.32	18.4	3.68	7.2
50	80	8.72	12.4	2.63	3.8
	90	8.64	12.5	2.98	4.3
	100	8.81	12.7	3.25	4.7

7.3.3 Transmission ratio

7.3.3.1 Resultant acceleration iliac : platform ratio

A significant main effect for frequency was found (F [1.27,11.39] = 809.60, $p < 0.001$ with near perfect effect size $r = 0.99$), Figure 7.9. Post hoc analysis revealed significant decreases in A_R I : P between: 30 and 40 Hz (mean difference 0.11, $p < 0.001$); 30 and 50 Hz (mean difference 0.18, $p < 0.001$); and 40 and 50 Hz (mean difference 0.07, $p < 0.001$). These results suggest that dampening between platform and iliac levels increased at higher frequencies.

No significant main effect for external load was found (F [2,18] = 0.50, $p = 0.61$ with small effect size $r = 0.22$). No significant frequency x external load interaction effect was found (F [2.03,18.29] = 0.36, $p = 0.70$ with small effect size $r = 0.20$).

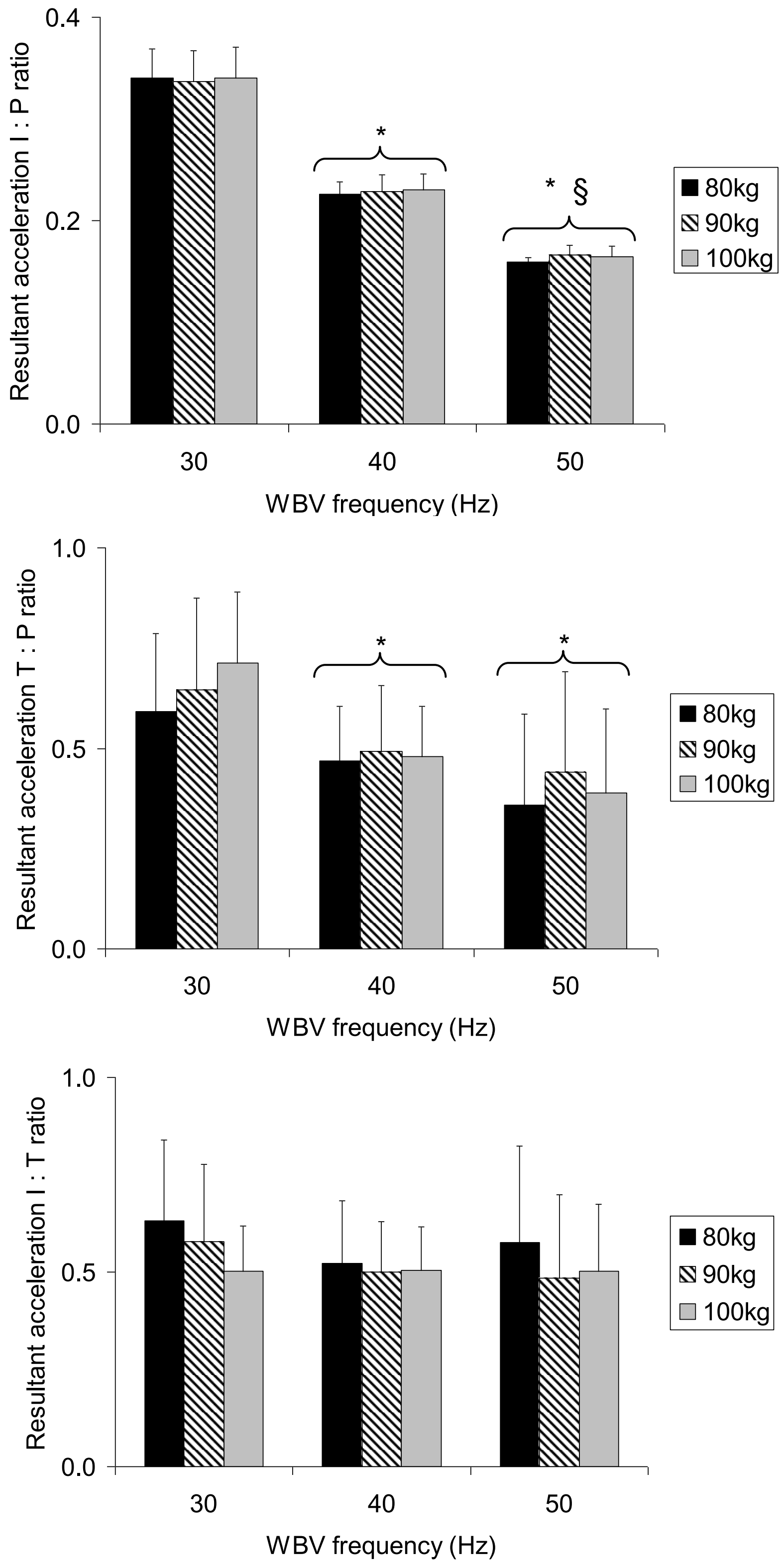
Figure 7.9: Mean ± SD peak A_R transmission ratios during three WBV frequencies and three external loads. A significant main effect for frequency was found for A_R I : P and A_R T : P.

Figure A, I : P ratio; Figure B, T : P ratio; Figure C, I : T ratio; *, significantly lower than 30 Hz §, significantly lower than 40 Hz.

A

B

C



7.3.3.2 Resultant acceleration tibia : platform ratio

As a significant main effect for frequency was found (*F* [1.15,10.03] = 9.53, $p < 0.05$ with large effect size $r = 0.71$), Figure 7.9. Post hoc analysis revealed significant decrease in A_R T : P between: 30 and 40 Hz (mean difference 0.17, $p < 0.05$); and 30 and 50 Hz (mean difference 0.25, $p < 0.05$). In contrast to A_R I : P, no significant difference between 40 and 50 Hz were found ($p = 0.22$). The significant decrease in A_R T : P ratio suggests an increase in lower leg dampening as WBV frequency increases. This is apparent between 30 and 40 Hz or 30 and 50 Hz. However, lower leg dampening does not significantly increase with a change of frequency from 40 to 50 Hz. These results suggest that dampening between platform and tibia increased at higher WBV frequency but not significantly between 40 and 50 Hz.

As for A_R I : P, no significant main effect for external load was found (*F* [1.27,11.40] = 3.20, $p = 0.09$ with large effect size $r = 0.51$). No significant frequency x external load interaction effect was found (*F* [1.75,15.72] = 1.61, $p = 0.23$ with moderate effect size $r = 0.39$).

7.3.3.3 Resultant acceleration iliac : tibia ratio

Unlike the previous transmission ratios, no significant main effect for frequency was found for A_R I : T (*F* [1.27,11.43] = 0.79, $p = 0.42$ with small effect size $r = 0.28$), Figure 7.9. This suggests frequency does not influence dampening between tibia and iliac levels.

Also in contrast to the previous transmission ratios, a significant main effect for external load was found (*F* [2,18] = 4.11, $p < 0.05$ with large effect size $r = 0.56$). However, post hoc analysis revealed no significant differences in A_R I : T between external loads. No significant frequency x external load was found (*F* [1.66,14.92] = 1.14, $p = 0.34$ with moderate effect size $r = 0.33$).

7.3.4 Relationship between resultant acceleration and electromyography

7.3.4.1 Relationship between resultant acceleration and electromyography

There were no significant correlations between A_R and EMG for the three WBV frequencies and three external loads ($p < 0.05$) for vastus lateralis and rectus femoris muscles (Pearson correlation $r = 0.07$ to 0.62). For both quadriceps muscles, during 50 Hz frequency at 100 kg external load the correlation was close to significance (p value = 0.054). Also during 50 Hz and at 100 kg one significant correlation between A_R and gastrocnemius EMG was found (Pearson correlation $r = 0.68$, $p < 0.05$).

7.3.4.2 Relationship between resultant acceleration iliac : platform and electromyography

No significant correlations across any frequency or external load were found between A_R I : P and vastus lateralis EMG (Pearson correlation r range of 0.03 to -0.45, $p > 0.05$). However, one significant correlation was found between A_R I : P and rectus femoris EMG during 50 Hz and 100 kg (Pearson correlation $r = -0.63$, $p < 0.05$)

7.3.4.3 Relationship between resultant acceleration tibia : platform and electromyography

Significant correlations between A_R T : P and gastrocnemius EMG were only found during 30 Hz at 80 kg and during 50 Hz at 100 kg (Pearson correlation $r = 0.75$, $p < 0.05$ and Pearson correlation $r = -0.63$, $p < 0.05$ respectively).

7.4 DISCUSSION

7.4.1 Electromyography

The main finding of this present study was that external load significantly influenced quadriceps, but not gastrocnemius, EMG activity during WBV. The higher the load on both quadriceps muscles the higher the EMG response, as expected (Clark et al., 2012). This was the case for all external loads for vastus lateralis and for most loads for rectus femoris. The other finding of this present study was that WBV frequency did not significantly affect quadriceps muscles EMG response in support of chapters 4 and 5. However, WBV significantly increased gastrocnemius EMG activity versus no WBV (0 Hz); although no difference between WBV frequencies (30, 40 and 50 Hz) was reported. Finally, when looked at group analysis, external load did not influence EMG response to varying WBV frequencies for any of the three muscle groups recorded; although there appears evidence of a possible individualised response from both external load and WBV frequencies.

The findings regarding the influence of external load on EMG support the long established consensus that muscle activation increases are directly proportional to external load (Clark et al., 2012). In this case the external load was in the form of artificially increased external loads via weighted vests. Specifically, the increased EMG activity in both quadriceps due to higher external loads supports previous work which reported significantly higher EMG activity levels during increased loaded squats (McCaw & Melrose, 1999). Although based on dynamic squatting and not isometric squatting, (as used by this present study), the illustration of increased EMG activity in those muscle groups considered important for squat biomechanics follows grounded physiological principles of resistance training (McCaw & Melrose, 1999). It would be highly likely the same principles, in the context of load, can be applied to isometric squatting of varying external loads. This also support Ritzmann *et al.* (2012) who reported increased quadriceps EMG activity with higher loads during an isometric squat.

The lack of influence of external load on gastrocnemius EMG activity supports the theory that during isometric squats the quadriceps are the main muscle group (McCaw & Melrose, 1999). Therefore, increased external load may have a lesser effect on shank musculature. The findings also support work by McBride *et al.* (2006) in which gastrocnemius EMG activity was not influenced by a change in external load; but did affect quadriceps muscles. However, the finding of this present study is in conflict with Ritzmann *et al.* (2012) who reported gastrocnemius EMG activity increased by 14 % with an increase in load of 33 % BM. Differences between the present study and the work by Ritzmann *et al.* (2012), including WBV platform type and ankle joint angle, are likely reasons to explain the conflict. The present study also conflicts with Hazell *et al.* (2010) who reported an increase in gastrocnemius EMG activity of 6 % following a similar addition of external load (30 % BM). While this increase in EMG activity was during WBV it was also during dynamic squatting not isometric as in the present study. The demands of dynamic squatting are likely reasons to explain why gastrocnemius EMG activity increased.

The addition of load is also a reason for the conflict as load was calculated as a percentage of BM; whereas in the present study each external load was standardised (80, 90, 100 kg). Taking the mean BM data (84 kg) of participants recruited by Hazell *et al.* (2010) the additional 30 % external load totalled approximately 110 kg. This is higher than the present study used and may explain why increases in gastrocnemius EMG activity were reported. The finding that gastrocnemius EMG activity was significantly higher during WBV versus no WBV supports Ritzmann *et al.* (2012) who reported gastrocnemius EMG was more frequency dependent than other muscles such as rectus femoris. The findings also support work utilising a similar WBV platform type and frequency (35 Hz) which reported increased gastrocnemius EMG activity during WBV versus no WBV (Roelants et al., 2006).

Post-hoc analysis of the main effect of external load for vastus lateralis and rectus femoris revealed differences between the muscle groups. Vastus lateralis illustrated differences in EMG response between all the three external loads; whereas rectus femoris exhibited only significant differences when comparing 80 kg with the other two external loads. The difference between the two muscle groups may likely be explained by the muscular articulations. Rectus femoris, as a bi-articulate muscle, not only crosses the knee but also the hip joint (Thompson & Floyd, 1998); whereas vastus lateralis only crosses the knee joint (uni-articulate). The involvement of a second joint, especially the hip, during WBV exposure while adopting an isometric squat posture may explain the difference. This second articulation is likely to influence the extensibility of rectus femoris in comparison to vastus lateralis (Thompson & Floyd, 1998). In addition, individual muscles within the quadriceps group play different roles with regards postural stability of the knee and patellar-femoral joints (Thompson & Floyd, 1998). It could be speculated that the difference in vastus lateralis response to 80, 90 and 100 kg loads may indicate a greater stability role during WBV exposure.

The finding of this present study that for the quadriceps muscle group, frequency of WBV did not influence EMG response supports the findings presented in chapter 5 where no frequency main effect was reported. As discussed in sections 5.4.1 and 5.4.3.1 this was across all participants as a group and at individual levels, evidence for an individualised response to WBV frequency may exist. As the mean BM of participants investigated in chapter 5 (mean = 82.6 kg) was less than the present study (90 and 100 kg), therefore, for heavier individuals, as a whole frequency does not appear to influence quadriceps EMG response. However, a similar individualised response to frequency may still exist for these heavier users of WBV.

The findings also support Petit *et al.* (2010) and Pollock *et al.* (2010) who both failed to find an effect of varying WBV frequency on quadriceps EMG activity. It should be noted that Petit *et al.* (2010) utilised EMG responses evoked during a twitch protocol (artificially evoked contractions via electrical stimulation) and the D_{PTP} values were simultaneously changed along with frequency. Therefore, it is unknown if frequency alone is responsible for the lack of influence on EMG reported. Work by Pollock *et al.* (2010) utilised lower frequencies WBV via a pivotal WBV platform type. It is of interest that, for higher frequencies, delivered from a different type of platform, frequency did not appear to influence EMG response at a group level. However, caution should be taken when comparing WBV studies as limitations exist in the work by Pollock *et al.* (2010) (e.g. low WBV frequencies used and posture adopted during WBV, see section 7.1). The findings do conflict with Cardinale & Lim (2003b) who reported 30 Hz elicited the highest vastus lateralis EMG response.

Reasons that may explain the conflict have been discussed in chapter 5 (section 5.4.1).

The present study reported no main effect for frequency for quadriceps EMG activity, including no difference compared to 0 Hz. This is in conflict with Moras *et al.* (2010) who reported EMG activity was significantly higher during vibration versus no vibration. However, this was based on exposure to vibration via a hand held bar and thus upper limb EMG was recorded. The present study also conflicts with Roelants *et al.* (2006) who reported quadriceps EMG activity was significantly higher during WBV versus no WBV. However, this was based only on 35 Hz and involved dynamic exercises performed on the WBV platform. The difference between that protocol and the isometric protocol completed in the present study may explain the conflict. It could be speculated that the lack of frequency effect on quadriceps EMG may be due to two additional reasons.

Firstly, the magnitude of WBV stimuli may not be sufficient to elicit a frequency specific response. This may support the overall lack of jump performance benefit presented in chapter 5. Secondly, the location of the muscle group in relation to the distance from the WBV platform may be a factor. The quadriceps are distal to the location where WBV stimulus is first exposed to the body (Ritzmann et al., 2012). As a consequence the transmission of WBV stimulus to the muscle group may be reduced (Pel et al., 2009). This will be discussed in section 7.4.4, but briefly; if the WBV stimulus has been dampened by musculature, tendons, ligaments and bone more proximal to the platform, then the EMG response to frequency may be reduced. This theory is given strength as calf EMG activity measured in the present study was significantly higher during WBV (30, 40 and 50 Hz) versus no WBV (0 Hz). This may suggest that when EMG is recorded in muscle groups more proximal to the platform then responses to WBV frequency may be detected.

An explanation of the contrast in WBV response of gastrocnemius versus quadriceps EMG activity may involve the location of these muscles in relation to the WBV platform. Previous work has suggested that frequency related increases in EMG activity may be muscle location dependent (Ritzmann et al., 2012). Obviously, the proximity of gastrocnemius versus quadriceps muscles to the WBV platform may explain why frequency effects for the former muscle were reported. Muscles which are closer to the WBV platform may be exposed to a higher magnitude of stimuli than at higher body segments. This will be discussed in section 7.4.4.

A further explanation may involve the instruction given to participants to distribute their weight predominantly on their forefoot, but allowing some heel contact with the WBV platform. Work by Ritzmann *et al.* (2012) illustrated that forefoot stance during WBV significantly increased calf EMG activity versus normal stance. The same work also reported that forefoot stance significantly reduced quadriceps EMG activity versus normal stance. This is extremely pertinent to not only this present study but also to chapter 5. By instructing participants to avoid excessive weight distribution through the heels, on safety considerations (to avoid high transmission of WBV to the torso and head, see section 2.6), it may have specifically targeted the WBV stimulus to calf musculature but not to quadriceps muscles. The posture adopted may be optimal for calf exposure to WBV but detrimental to quadriceps exposure (Ritzmann et al., 2012), hence the frequency dependent finding of this present study.

The combination of both foot stance and posture may account for evidence of an individualised response to both external loads and WBV frequencies in this present study. These individual responses in muscle activity of all three muscle groups may indicate that each participant is reacting to the external load and WBV frequency differently. This would support works by Di Giminiani (2009; 2010) who reported varying EMG responses in participants to the same WBV frequency. Inter-participant posture may account for this; but it may also be due to participant specific characteristics, as measures of neuromuscular excitability (H-reflex) have been reported to be individual (Armstrong *et al.*, 2008). However, it should be acknowledged that the aim of this present study was not to investigate possible reasons for an individualised response, and further research should be conducted.

In conclusion, the influence of external loading on quadriceps EMG activity during WBV was as expected. Although differences in main effects of external loading and frequency exist when comparing quadriceps and calf EMG activity. This is likely due to both specific biomechanics of the squat posture adopted and the distance from the WBV platform and subsequent transmission of the WBV stimulus to each muscle group. The use of accelerometers can quantify the magnitude of stimuli at muscle groups and will be discussed in the following section.

7.4.2 Resultant acceleration

The main finding of this present study was that external load did not influence A_R at platform level, this was regardless of WBV frequency. The other finding was that frequency significantly increased A_R at platform level, as expected. This was true for all WBV frequencies utilised, but only at platform level, not at higher body segments. A third finding was that A_R was significantly lower at superior body segment from the WBV platform. This will be further discussed in section 7.4.4.

The finding that external load did not influence A_R supports Pel *et al.* (2009) who reported that, for a tri-planar WBV platform type, external load did not appear to influence acceleration. This was based on an acceleration change of less than 10 % during WBV involving 62 and 82 kg participants versus unloaded WBV. It should be noted that, while the WBV frequencies were similar to this present study, the platform type was different; affecting direct comparison between the two works. Also important, was the use of lower body masses (62 and 82 kg) than the present study (80, 90 and 100 kg). Pel *et al.* (2009) did utilise a 100 kg BM participant but with a different WBV platform, described as a horizontal oscillation type. In this example, Az increased by nearly eight fold, while reducing Ax and Ay. This highlights the problematic issues with comparisons across WBV platform types. The findings by Pel *et al.* (2009) that 100 kg load significantly alters acceleration may appear to conflict with this present study; however, this needs to be viewed in context. The horizontal oscillating direction may be more susceptible to changes due to additional external load applied to the platform by the participant standing on the platform. It should also be noted that acceleration data was collected as tri-axial but not presented as A_R as per this present study. It does appear that the present study finding advances the existing literature of Pel *et al.* (2009) to include heavier external loads (90 and 100 kg).

With regards the finding relating to frequency proportionally influencing A_R, this is in support of the approaches taken by Rauch *et al.* (2010) and Wilcock *et al.* (2009) (see section 2.1) and supports findings by Pollock *et al.* (2010) and Preatoni *et al.* (2012). Finally, the lower A_R recorded at higher body segments, also supports Cook *et al.* (2011), which will be further discussed in section 7.4.4.

There are several explanations that can account for the findings of this present study, in terms of A_R. The minimal influence of external load appears to have on platform A_R suggests that the WBV platform output remains stable. This is regardless of whether a user was 80, 90 or 100 kg and also whether 30, 40 or 50 Hz frequencies were utilised. It would appear that the machine mechanics can tolerate heavier loads, consistently delivering a set magnitude of WBV stimuli. However, this is only relative to other WBV frequencies and it could be argued that without an external reference point all three frequency outputs may have been equally affected by external load. This aspect will be discussed in chapter 8, as the use of accelerometers and the mathematical approaches of Rauch *et al.* (2010) and Wilcock *et al.* (2009) allow direct comparison between theoretical and actual magnitudes of WBV parameters and thus stimuli.

However, as WBV stimuli were transmitted through musculoskeletal structures, the same proportional relationship between frequency and acceleration was not found. This is highly likely to be due to dampening of the WBV stimulus. This may be via passive mechanisms as the stimuli passes through skeletal structures (Rubin *et al.*, 2003; Kiiski *et al.*, 2008); or via active mechanisms whilst passing through muscular structures (Crewther *et al.*, 2004; Cook *et al.*, 2011). To support this, the present study also found an inversely proportional relationship between A_R and the distance from platform (i.e. higher body segments), which will be discussed in section 7.4.4.

The present study demonstrated a dampening effect at higher body segments; and identified a potential method for quantifying WBV stimuli at specific body segments, including a potential dose-response model (Crewther et al., 2004). It also confirmed very low magnitudes of WBV stimuli transmitted to the hip, which likely means equally low transmission of WBV to the torso and head, a reassuring finding regarding safety considerations (avoiding high magnitudes of WBV at these segments, see section 2.6) (Mester *et al.*, 2006; Pollock *et al.*, 2010). However, as no accelerometer was placed in the torso and head body segments this remains speculative.

In conclusion, A_R appears resistant to external load and increases as expected with WBV frequency, furthering the WBV literature utilising heavier external loads and higher WBV frequencies. There appears a dampening mechanism that would account for reduced A_R at higher body segments. Obtaining A_R for specific levels allows a potential dose-response model to be developed, but it also confirms a very low probability of unsafe WBV transmission to the torso and head.

7.4.3 Individual axis acceleration

The main finding of this study was that, for Az, the pattern of results largely replicated A_R. There were two exceptions to this: one was a significant main effect for external load; however, no post hoc differences were detected, suggesting a possible type I error; the second difference was that external load appeared to influence Az differently (during 30 and 50 Hz WBV only), depending what level the acceleration data was recorded at, (i.e. platform, shank or thigh). On further analysis, Az significantly decreased as height of body segment increased, for all external loads, again during 30 and 50 Hz only. The final finding was that horizontal accelerations contributed approximately a third of total A_R values and that, at lower WBV frequencies, both Ax and Ay accounted for a greater proportion of A_R than at higher frequencies.

To the author's knowledge this is the first study to quantify tri-axial accelerations from a vertical oscillating WBV platform. Pel *et al.* (2009) investigated tri-axial accelerations at platform level using a tri-planar WBV type. The finding by Pel *et al.* (2009), that horizontal accelerations were relatively minimal in relation to A*z* appears to conflict with the present study. However, Pel and co-workers reported accelerations of an unloaded platform. The study did not present A_R data, although by visual inspection of Pel *et al.* (2009) data A*z* represented a very large proportion of the total A_R (≈ 90 %). It could be speculated that unloaded platforms may perform differently. Utilising three external loads may have induced higher horizontal accelerations (both A*y* and A*x*). This may explain the conflict between the present study and Pel *et al.* (2009).

As the major frequency during heel strike is 10 – 20 Hz (Nigg & Wakeling, 2001), it may be speculated that 30 Hz WBV would match this the closest, resulting in higher muscle activation and potentially higher dampening (Nigg & Wakeling, 2001). Therefore, at 30 Hz WBV A*z* transmission may have reduced, and be more susceptible to the influence of external load. This is an area which requires further research, it is important to highlight that A_R was not found to be external load dependent and did not exhibit any accelerometer location and external load interactions. This is relevant as A_R represents the total acceleration exposed to WBV users.

The final finding presented in this section relates to the inverse relationship between WBV frequency and the percentage of A_R output that were in the horizontal axes. As frequency increased, the contribution of horizontal accelerations to A_R reduced, which appears to suggest that as the platform motor generates a faster oscillation a greater proportion is within the vertical axis. As this platform is manufactured as a vertical WBV platform it appears that at lower frequencies the platform may not be performing effectively. It is also interesting to note that at 30 Hz external load increased the percentage of horizontal acceleration contributions towards A_R (38, 40 and 43 % for external loads 80, 90 and 100 kg respectively). It appears that as external load increases the performance of the WBV platform may deteriorate away from delivering the specified vertical WBV stimuli. Overall the magnitude of A_R may not change with altered external load. However, the individual tri-axial components that make up A_R may be influenced by external loading. The effect of which on physiological responses is largely unknown.

7.4.4 Transmission ratio

The main finding of this study in relation to transmission ratio was that A_R I : P and A_R T : P ratios were significantly reduced at higher WBV frequencies. The most marked decreases were seen in A_R I : P ratio at which all frequencies were significantly different. Whereas, A_R T : P ratios were only reduced when comparing 30 Hz with other frequencies (40 and 50 Hz). A reduced transmission ratio implies a greater magnitude of dampening between locations, characterised by lower A_R at higher body segments. Therefore, it appears dampening of WBV stimuli between platform and iliac and between platform and tibia locations increases during WBV of higher frequencies. Finally, it appears external load does not influence the dampening mechanism; as, similar decreases in vibration transmission occurred for WBV users of 80, 90 or 100 kg BM.

The findings support Crewther *et al.* (2004) and Cook *et al.* (2011) who reported WBV frequencies above 20 Hz would be associated with active dampening to reduce WBV stimuli transmission; and increased dampening at higher WBV frequencies respectively. This would account for the lower acceleration magnitudes recorded at higher body segment sites as reported by Crewther and co-workers and by this present study.

The reported 66 to 84 % reduction in WBV stimuli at knee and hip levels respectively, are lower than Pel *et al.* (2009) (91 and 97 % respectively). The reason for the difference in reduction of WBV stimuli is likely due to: different WBV platform (tri-planar versus vertical respectively); and differences in WBV frequencies utilised (25 Hz versus 30 to 50 Hz). The present study reported increased dampening at lower WBV frequencies, suggesting the frequency of 25 Hz would evoke higher dampening mechanisms and thus a higher percentage of WBV reduction. The reduction of WBV stimuli appears greater than Pollock *et al.* (2010). Again differences in WBV platform type should be highlighted as WBV frequency was limited to 30 Hz compared to 30 – 50 Hz in the present study.

Following the finding of significant decreases in A_R at higher body segments (see section 7.4.2), a dampening mechanism seems plausible. This mechanism appears to increase during higher WBV frequencies, as demonstrated by lower transmission ratios. Dampening mechanisms are likely to be associated to a muscle tuning theory (see section 2.3.6). Wakeling *et al.* (2002) suggested a muscular response dependent on the excitation frequency closely matching a natural resonant frequency. As the leg musculature was already in an active state during 90 º knee flexion squat, the WBV stimuli is likely to have been absorbed via a detachment and cross bridge cycling mechanism (Wakeling et al., 2002). The musculature in both lower and upper leg segments would dissipate and dampen the WBV stimuli (Cardinale & Wakeling, 2005). This would be characterised by lower A_R transmission ratios. The location of the accelerometers in the present study allows whole leg, and individually upper and lower leg A_R magnitudes and dampening to be quantified.

With regards A_R T : P (a measure of dampening in the lower leg) as WBV frequency increased so too did the dampening. This was true for 30 versus 40 Hz and 30 versus 50 Hz. This could be accounted for by a higher response in EMG activity thus dampening the WBV stimuli transmission between platform and tibia locations. The muscle group that is likely to be a prominent role in potentially dampening WBV stimuli would be the calf musculature. As the EMG activity for gastrocnemius was frequency dependent (see section 7.4.1) the increased dampening occurring at higher frequencies is likely to be through higher muscular activity. As WBV frequency increased, the magnitude of A_R at platform level also increased; as a result, calf EMG activity increased and consequently so too did dampening capacity.

A similar mechanism can be put forward for A_R I : P transmission ratio as this is likely to measure the dampening capacity of the entire lower limb body segment.

This obviously includes the lower leg and, as shown in section 7.3.3.1, a 35 to 61 % reduction in A_R occurs between platform and tibia. Therefore, it could be speculated that the component involving platform to tibia dampening could account for a significant portion of the dampening mechanisms reported for the total lower limb.

This is supported by the finding that A_R I : T did not decrease during higher WBV frequencies. This particular ratio may quantify the dampening in the upper leg. As this dampening did not increase at higher frequencies it could be that active dampening, via increased EMG activity, may play a lesser role specifically in the upper leg. This speculation is strengthened by the report that quadriceps EMG activity was not frequency dependent (see section 7.4.1). Therefore, the A_R dampening between tibia and iliac reported may not be attributed to active dampening via increases in quadriceps EMG activity. If not through active dampening then passive dampening is another candidate (Rubin et al., 2003). This would involve reduction of the WBV stimuli as it was transmitted through structures such as bone, cartilage and joints. Obviously, during transmission from platform to iliac levels, the WBV stimuli will pass through a higher quantity of these structures. As a consequence, the majority of the loss in A_R when recorded at iliac versus tibia levels may be attributed to passive dampening.

Finally, the impact of limb posture should be highlighted. While adopting the squat posture the lower leg remains largely within the vertical axis. Whereas, the upper leg is in a horizontal axis, due to the 90 º knee angle. As the WBV platform produces a vertical direction of oscillation, this could explain the differences discussed between WBV transmission in the upper and lower leg segments. Other factors associated with transmissibility and specific body segments may also account for the difference, including the musculature volume of each segment (Kiiski et al., 2008); differences in muscle roles adopted in a squat posture during WBV, (proprioceptive versus strength role during 90 ° squatting).

The use of accelerometers on different body segments is a methodological strength in terms of obtaining valuable data on a potential dampening mechanism and also a dose – response relationship. However, by using the skin mounted approach, the present study should acknowledge possible limitations. Previous studies utilising skin mounted accelerometers have conceded potential limitations using this approach, such as skin movement errors (Kiiski *et al.*, 2008; Pel *et al.*, 2009). Even so, this type can provide valuable acceleration data during WBV (Crewther *et al.*, 2004; Kiiski *et al.*, 2008; Fratini *et al.*, 2009a; Pel *et al.*, 2009). The advantage of the skin mounted approach is the non-invasive nature however; disadvantages include potential errors occurring through skin movement (Matsumoto & Griffin, 1998; Pel *et al.*, 2009). Furthermore, large inter-subject variability in vibration transmissibility from floor to ankle and hip levels has been reported (Harazin & Grzesik, 1998).

Correction factors have been developed for the skin mounted approach (Kitazaki & Griffin, 1995) however, these were not developed in the context of WBV as a platform use. Previous studies utilising skin mounted accelerometers in this context have not performed correction factors (Crewther *et al.*, 2004; Kiiski *et al.*, 2008; Pel *et al.*, 2009; Cook *et al.*, 2011). This may be surprising as it has been suggested that without correction, skin mounted accelerometers may overestimate acceleration values by 10 – 20 % (Kiiski et al., 2008). An alternative to skin mounted accelerometers is a bone mounted approach. This has significant methodological impact in terms of: participant recruitment in relation to fixation of accelerometers onto bony landmarks; and justification for this approach in trained athletes exposing them to increased infection risk and impacting their training. The use of accelerometers whether skin or bone mounted is relatively new in terms of WBV research. Therefore, further research is required to investigate and validate an appropriate use of accelerometers when attempting to quantify WBV stimuli.

An additional limitation is the potential that, due to low participant numbers, the study may be underpowered and fail to identify significant differences in not only transmission ratio, but previously discussed variables also.

To summarise, a lower transmission ratio between platform and higher body segments, suggests that a dampening mechanism increases during higher WBV frequencies. The dampening phenomenon is likely to be due to increased muscular activity of the lower leg to dissipate the WBV stimuli. This is especially true for the lower leg as EMG activity matched a similar WBV frequency dependent relationship. However, with regards to the upper leg body segment no such frequency dependent relationship was reported for quadriceps muscle activity and therefore, passive dampening may play a significant role of dampening of WBV stimuli. The increased distance from the WBV platform to higher body segments, and thus the greater magnitude of structures the WBV stimuli is transmitted through suggests that passive dampening may play more of a role. There may also be a positional aspect of leg segments to dampening in relation to the vertical oscillation direction of this particular WBV stimulus.

7.4.5 Conclusions

The aim of this study was to investigate the influence of external loading on EMG responses and acceleration magnitudes during acute WBV. External load influenced quadriceps EMG activity as expected, although calf EMG activity was unaffected by load. WBV frequency did not influence EMG responses in support of chapters 4 and 5. However, for calf EMG, WBV increased activity versus no WBV. The differences in EMG responses between muscle groups are likely due to the biomechanical demands of the squat posture adopted; and due to the distance from the WBV platform and therefore subsequent transmission of WBV stimuli.

External load did not influence A_R at platform level regardless of WBV frequency and, as expected, greater A_R was recorded at higher frequencies. Therefore, external load appears not to reduce the WBV magnitude across relative frequencies. However, differences in actual WBV stimuli recorded via tri-axial accelerometers included a larger than expected component of A_R within the horizontal, (medial-lateral) axis. This highlights the rationale why WBV stimuli should be measured and assessed for accuracy, as previously undiscovered characteristics of the WBV stimuli were identified (in addition, see discussion relating to recorded versus theoretical D_{PTP} magnitudes section 8.4).

The use of tri-axial accelerometers also allowed for the identification of a potential dampening mechanism at higher body segments. This dampening increased during higher WBV frequencies and was characterised by lower A_R at both lower leg and upper leg body segments. This is likely due to increased muscle activity which dampens the WBV stimuli, especially in the gastrocnemius muscle. Low magnitudes of A_R were also reported at the pelvis level, providing confirmation of low transmission of WBV stimuli to the torso and likely head areas.

Chapter 8: Discussion

The aim of this thesis was to characterise both the neuromuscular responses to acute WBV but also the WBV stimulus itself. In both elite and trained athletes it appears there may be an individualised response in both EMG activity and performance measures to varying WBV frequencies. This has ramifications across the research as well as the application of WBV as a training stimulus. Unless each individual is assessed for their possible individualised frequency then subsequent protocols will likely be less effective. A method to achieve this could be via EMG analysis and this thesis has put forward new rationale demonstrating that filter technique can significantly affect EMG data recorded during WBV. This thesis has also proposed a possible filter technique which may exclude WBV platform noise but not true muscle signal.

The thesis has also presented valuable insight into a typical WBV stimulus, confirming good accuracy of frequency and overall A_R appears resistant to external loads (of up to 100 kg BM). However, D_{PTP} was different than manufacturer's specifications (see section 8.4). The "purely vertical" oscillating WBV stimulus unexpectedly contained horizontal components of acceleration. This highlights the need to verify WBV parameters using methods and designs presented in this thesis. This is especially important as there is evidence that WBV responses are dependent on the direction of oscillation. The use of accelerometers placed on different body segments suggests a new quantifiable dampening physiological response which appears to increase during higher WBV frequencies. This may allow further insight into mechanistic responses to WBV, but also a method of monitoring g-force magnitude in relation to safety considerations. The vast majority of WBV literature would benefit from this method to determine safe protocols, a vital aspect of any new training method.

8.1 The response of electromyography to whole body vibration

This thesis began investigating the neuromuscular responses to WBV in elite athletes, reporting that not one single frequency consistently elicited a higher EMG response. The findings presented in chapter 4 also suggested an individualised response to WBV frequency. There were those participants who responded with large EMG activity at high WBV frequencies; those who responded during low frequencies; and those who failed to illustrate a WBV response. It has been suggested that the reflexive nature of the EMG response may be frequency dependent (Pollock et al., 2012). The work presented in chapters 4 and 5 did not support this. A likely reason for this is the difference between vibration stimulus applied directly to a muscle tendon (as performed by Eklund & Hagbarth) and WBV stimulus (as delivered in this thesis). In addition, greater frequencies of vibration were applied directly to the muscle tendon.

Data does suggest there is an EMG individualised response but this is not conclusive. The implications to WBV are wide-ranging across the literature. This may, to some extent, account for the conflicting findings reported in the literature; as the investigations have largely utilised one set frequency across all participants. Therefore, leading to speculation that only a small proportion may have been receiving their "individualised" WBV frequency and thus only those participants may have responded positively. The exact mechanisms to explain why an individual would respond to a certain WBV frequency over another remain largely unknown. One possible reason involved the influence of external load on the WBV output delivered by the platform. This was especially pertinent as participants in chapter 4 were in excess of 100 kg BM. The influence of external load on WBV magnitude was the focus of chapter 7 and will be discussed further in section 8.4. One of the main findings from chapter 7 suggested an individualised ability to dampen WBV stimuli, which adds support to the argument of a participant-specific neuromuscular response to WBV. The dampening capacity requires further research and will also be discussed in section 8.5. The nature of the data collection completed in chapter 4 meant limitations existed, including the lack of performance measures. This influenced the rationale of study design in chapter 5 to include pre and post WBV jump performance outcome measures. This will be discussed in section 8.3. Another limitation identified was the short WBV duration due to the availability of participants, which may have influenced the true EMG response to WBV frequency. Again, this informed the study design for chapter 5. The finding of that chapter supported an individualised EMG response to frequency, even at longer WBV durations (60 s).

By the very nature of research across participants as a group, statistical analysis may be inefficient in detecting individualised responses. Reasons to account for the specific responses are still to be eluded. This is due to the fact research into this aspect of neuromuscular responses to WBV is still emerging. Only two studies (Di Giminiani *et al.*, 2009; Di Giminiani *et al.*, 2010) have presented data directly comparing individualised EMG responses in active participants only. The findings of chapter 4 and 5 suggest a similar individualised response exists in elite, highly-trained athletes. Reasons can be speculated here which include: differences in muscle fibre characteristics, based on pre-existing H-reflex individualised responses (Armstrong et al., 2008); individual sensitivity to WBV (Di Giminiani et al., 2010) via specific threshold levels in reflex activity which may govern the neuromuscular response to WBV (Cardinale & Lim, 2003a); differences in mechanoreceptors quantity and location (Pollock et al., 2012). Until these mechanisms are investigated, speculation will continue and further research is warranted.

Whether performance enhancing effects may result from this individualised EMG response is an important aspect to discuss and one which relates to the link between EMG activity and muscle force. This area of research has produced conflicting findings as both positive and negative correlations have been reported (Gerdle *et al.*, 1991) *cf.* (Komi & Vitasalo, 1976). This is reflective of the lack of general consensus in the literature, evidence supporting the link between EMG activity and performance measures has been reported (Karlsson & Gerdle, 2001); however this was based on specific signal analysis techniques (specifically different time-frequency methods).

There is evidence to support the link between EMG activity recorded during isometric contractions and performance measures (Onishi *et al.*, 2000) which is the type of global muscle contraction performed while standing on a WBV platform. Although based on wire electrodes recordings, not surface EMG. It could be argued that EMG offers an appropriate indication of muscular force generating capacity and thus linked with performance capacity. A specific advantage of EMG analysis with regards WBV is that the response to frequency for example can be assessed during the WBV stimuli. There has been initial evidence submitted that a more effective EMG activity response can lead to performance enhancements (increased jump height) following chronic WBV (8 weeks) (Di Giminiani *et al.*, 2009). Therefore further research is required investigating the link between acute WBV responses in EMG and performance measures.

8.2 Electromyography data recorded during whole body vibration

Such further research involving EMG recording during WBV may prove problematic. An inherent difficulty in recording EMG via surface electrodes, which are sensitive to electrical signals, is the electromagnetic noise generated by the WBV platform. This thesis was aware of such electromagnetic noise signal in data presented in chapters 4 and 5 and thus designed an investigation into the influence of filtering strategies to minimise the impact of such noise, as previous studies have eluded to, but not directly investigated (Abercromby *et al.*, 2007a; Fratini *et al.*, 2009a; Fratini *et al.*, 2009b; Marin *et al.*, 2011). The fact that filter technique varies so widely across the WBV literature highlights the need for specific research into establishing the influence of filtering on EMG data collected during WBV. The lack of rationale for the application of a filter highlights this further. As demonstrated in chapter 6 the influence of filter technique is significant. This was in EMG data collected from commonly used WBV frequencies (30 – 50 Hz) which develops previous research, who reported at lower frequencies (5 – 30 Hz) electromagnetic noise within EMG data was minimal (Ritzmann et al., 2010). At higher frequencies, the impact of noise is likely to be greater as specific peaks within power spectra frequencies lay within “true” muscle signal ranges. Therefore, the choice of filter technique becomes even more crucial.

The results presented in chapter 6 illustrate the accurate removal of significant portions of EMG signal by applying filters. What remained to be determined was the composition of this removed EMG signal. By utilising a novel approach involving comparing filtered EMG data collected during WBV to “clean” EMG data collected during 0 Hz it can be proposed that the signal removed via notch filtering contained small portions of “true” muscle signal. Therefore, it is this thesis’s recommendation that a notch filter centred on the fundamental WBV frequency and its sub-harmonics, should be used in addition to the commonly utilised band pass filter technique. However, this was based on EMG data collected during vertical WBV, and thus further research should replicate the filter techniques on EMG data collected during WBV delivered by other platform types.

8.3 The response of jump performance to whole body vibration

One of the findings presented in chapters 4 and 5 was an individualised EMG response to WBV frequency. Previous research (Di Giminiani et al., 2009) reported a significantly greater positive response of jump performance following individualised WBV frequencies. One limiting factor of the work was the use of only active, but un-trained participants. Therefore, further research was warranted utilising well-trained. This was met in chapter 4; however, no performance measures were recorded; subsequently influencing chapter 5 which investigated the effect of WBV frequency on jump performance across well-trained participants. The main finding of that work was that frequency did not influence jump performance across participants as a group, indeed performance following WBV was not significantly different from 0 Hz.

There are several reasons that could account for this: firstly, the statistical approach as mentioned in section 8.1 may be inefficient when analysing, at group level, the existence of possible individual level responses. Secondly, the WBV parameters

(30 – 50 Hz; 3 mm D_{PTP}; and one repetition of 60 s) may provide too low a magnitude of WBV stimulus. This may be especially relevant in those well-trained participants utilised in chapter 5, as these individuals may have attenuated WBV responses (Marin & Rhea, 2010b). This is also relevant as the D_{PTP} value was taken from manufacturer's specifications. The study design in chapter 7 was such that, it allowed the magnitude of WBV parameters to be quantified and will be discussed in section 8.4. Thirdly, the influence of external loading on WBV output was speculated in chapter 5. However chapter 8 suggested that BM did not influence A_R output, but did present WBV stimulus data that could account for the lack of overall jump performance response across participants (section 8.4).

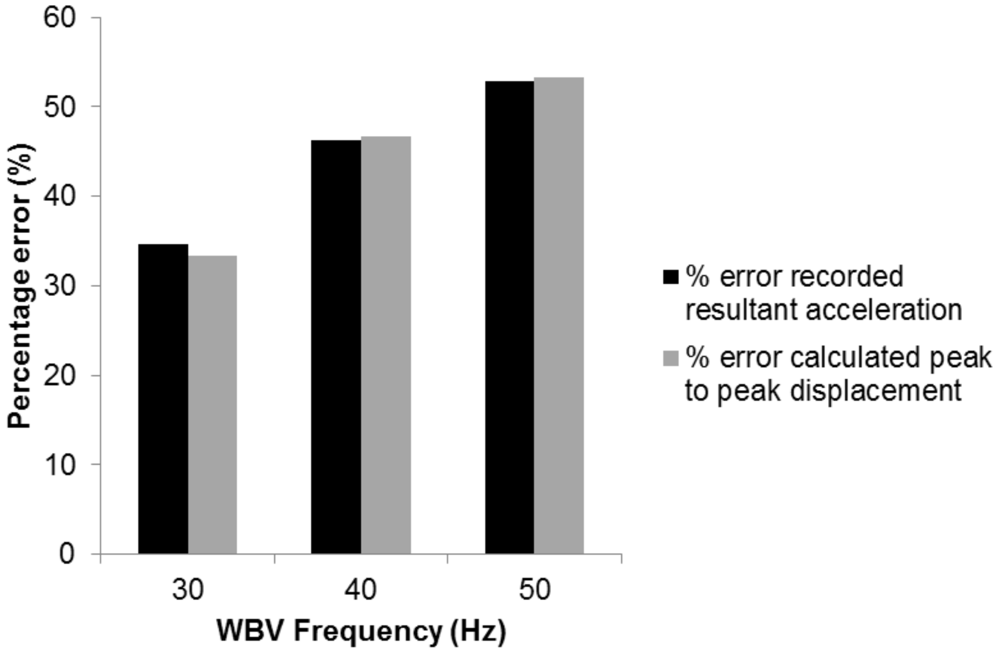
Similarly to EMG, individual responses in jump performance measures to WBV frequency were evident. Importantly, there was not match between the highest EMG during WBV and the biggest change in CMJ performance parameters. As discussed in sections 4.4 and 8.1 there are several reasons that may account for this, and future research should aim to establish whether an individualised response is present from other WBV platform types; and directly evaluate possible mechanisms.

The lack of volume-matched control groups has limited previous WBV research; as well as the lack of addressing placebo issues and possible role of expectancy effects (section 5.1). Little prominence has been afforded to this phenomenon, despite the potential to impact on post WBV performance (McClung & Collins, 2007). The results from chapter 5 suggest a minimal role of expectancy effect in this context. This is reassuring in relation to previous literature; however, the context of the findings should be highlighted. The role of expectancy may be minimal, but this is in acute settings, delivered by vertical WBV, and assessed via CMJ and DJ performance only. Application of these findings across WBV literature involving a wide range of parameters and platform types may not be appropriate. As the findings of chapter 5 also reported no overall jump performance response, the role of expectancy effect may be minimal due to that fact. Therefore, research which has elicited an overall positive influence on jump performance should be replicated and the role of expectancy using the study design of chapter 5 applied.

8.4 Quantifying whole body vibration and the influence of external load

One of the common themes from discussion sections in chapters 4 and 5 was the potential for external load to reduce WBV output and therefore the magnitude of WBV stimuli received by participants. External load did not appear to influence A_R across loads of 80, 90 and 100 kg, during 30, 40, or 50 Hz WBV frequencies. Again, context is required as these findings are relevant to vertical WBV and comparisons to other platform types, body postures and populations may be inaccurate. The methodology to determine the A_R output via accelerometers also allowed the frequency and D_{PTP} to be quantified. As the frequency output was accurate (see methodology section 3.3.2); and A_R was recorded, D_{PTP} was quantified (Rauch et al., 2010) and validated against the value provided in manufacturer's specifications (3 mm). Percentage errors of recorded A_R and calculated D_{PTP} under different WBV frequencies over all external loads are presented in figure 8.1.

Figure 8.1: Percentage errors of recorded A_R and calculated D_{PTP} under different WBV frequencies over all external loads.



As shown percentage errors of calculated D_{PTP} were large and increased with higher WBV frequencies. The D_{PTP} was not 3 mm, as per manufacturer specifications;

therefore, the actual magnitude of WBV stimuli exposed to participants was lower than the theoretical magnitude. Discussion points raised in sections 5.4.2 and in 8.3 are especially relevant as it was speculated that WBV parameters of 30 to 50 Hz and 3 mm D_{PTP} may provide an insufficient WBV magnitude to evoke a beneficial jump performance response across participants as a group. The difference between theoretical and actual D_{PTP} values support those reported by Preatoni *et al.* (2012) assessing a different manufacturer of vertical WBV platform type. A specified D_{PTP} was 4 mm but was found to range from 2.6 to 4.4 mm. Therefore, the potential for frequency dependency is confounded by changing D_{PTP}.

This present thesis has reported a difference of 1 to 1.6 mm in D_{PTP} values. This is likely to impact significantly on the A_R magnitude delivered to participants. Using the D_{PTP} value of 3 mm the A_R exposed to users during 30, 40 and 50 Hz WBV would be 53.3, 94.7 and 148 ms^{-2} respectively (see section 3.3.2). As reported, percentage errors of the calculated A_R values were between 33 and 53 %. Therefore a difference of 1 mm between specified (3 mm) and actual (2 mm) D_{PTP} is likely to be significant.

What was also apparent by this approach was that D_{PTP} values decreased with increasing WBV frequency. At 30 Hz, D_{PTP} was 2 mm but decreased to 1.6 and 1.4 mm during 40 and 50 Hz WBV respectively. Again, even a difference of 0.4 and 0.6 mm is likely to be significant as that corresponds to a 20 and 32 % decrease in A_R values respectively. Without dismantling the WBV platform, the reasons why D_{PTP} would decrease at higher frequencies is based on speculation. It may be that as the motor output increases in speed during higher frequencies, the transmission of oscillation between the motor and the platform surface may reduce. Therefore, the higher the frequency the greater the loss of D_{PTP} between motor and platform surface. Another reason could lie in the material of the platform surface itself. The motors which generate the oscillatory stimuli are housed internally. The stiffness or dampening coefficient of that material is unknown, and it may be reasonable to speculate that a degree of dampening may have occurred. This dampening may explain the difference between D_{PTP} values supplied by the manufacturer and those recorded on the platform surface. As WBV users stand at that location differences between stated and actual WBV magnitudes delivered to the user would be significant. If an inherent degree of dampening occurs between motor and platform, this should be accounted for, to maintain correct WBV parameters at the point of exposure to WBV stimuli. The finding of this thesis supports Rauch *et al.* (2010) and recommends that a validation procedure be completed before using a WBV platform in neuromuscular performance research. This can be done with the use of accelerometers using the methods and designs presented in this thesis (chapter 7).

8.5 Transmission of whole body vibration and the neuromuscular response

The use of several accelerometers placed at different body segments allowed WBV transmission to be characterised. The use of transmission ratios is relatively new with regards to WBV literature. The findings reported in chapter 7 suggest that dampening of WBV stimuli increases during higher frequencies. Therefore, it appears the capacity to dampen WBV stimuli increases depending on the magnitude of WBV stimuli. The EMG data of a lower leg body segment (i.e. calf EMG activity) supports this as it was higher during WBV versus 0 Hz. This was not the case for quadriceps EMG activity, suggesting a difference in the relationship between neuromuscular responses to WBV and the distance from the platform (i.e. higher body segments). This is likely associated with two types of dampening; active and passive (Rubin et al., 2003).

It is the active dampening which may characterise EMG activity. Both chapters 5 and 7 data suggested that the neuromuscular response may differ from other physiological responses. This is due to a minimal relationship between the individualised WBV frequencies as determined by EMG response versus that determined by jump performance responses. Both EMG and A_R responses to external loading during WBV differed in chapter 7. Perhaps the sole use of either EMG or explosive performance should be discouraged and both forms be utilised to gain a fuller understanding of WBV responses and potential individualised mechanisms.

The data in chapter 7 also confirmed a method to quantify the WBV magnitude at body segment levels. Although possible limitations exist regarding a skin mounted approach; nevertheless future research could utilise this methodology. This could include targeting WBV stimuli to specific body segments, via the adjustment of WBV parameter magnitudes or posture adopted. The alternative research avenue is the use to monitor acceleration magnitudes to the torso and head, reducing the risk of potential negative impacts (section 2.6). In addition safe and effective WBV protocols could be developed.

8.6 Practical applications

This thesis has provided evidence to suggest an individualised response to WBV frequency may occur in both EMG and jump performance measures. Users of WBV, researchers and other sport professionals need to be aware that individual athletes may respond differently depending on the WBV parameters chosen. A 'one size fits all' approach is likely to be ineffective at achieving optimal responses following WBV. One method to analyse a potential individualised response would be to utilise EMG recorded during varying WBV parameters. Based on evidence provided by this thesis, that electromagnetic noise / artifacts are unique to EMG recordings taken during WBV, EMG analysis has potential limitations. However, this thesis has presented evidence to support an effective filtering technique and offers rationale for the use of notch plus band pass filters. Notch filtering techniques offer an accurate and valid method of excluding WBV artifact signal from true muscle signal. It offers a practical solution to a problem identified by this thesis and by previous research when EMG activity is recorded during WBV. It appears that the role of expectancy effect may be minimal with regards this particular WBV stimulus during the protocol presented in this thesis. Researchers may be less concerned with possible placebo type influences in WBV study design. However, it is applicable for this WBV platform type only.

The use of multiple accelerometers across different body segments and on the WBV platform itself offers a practical method to monitor and quantify the magnitude of stimuli and possibly a dampening phenomenon. The WBV parameters utilised in this thesis resulted in very little additional accelerations being transmitted to the hip and therefore, it is unlikely that higher accelerations were transmitted to the torso and head. This thesis offers practical solutions and evidence to minimising potential negative effects of WBV stimuli reaching those body regions. By utilising multiple accelerometers the work presented here can give a practical suggestion that, for this WBV platform type, different parameters may be required depending on whether lower or upper leg segments, (and therefore muscle groups), are being targeted for WBV exposure.

It appears that a potential dampening factor (varying external load on the WBV platform) exerts minimal impact on the WBV output; and therefore for this particular WBV platform type; users of 80 – 100 kg can be confident that additional load will not have an additive detrimental effect. However, this should only be after validation of the WBV output itself has occurred (see recommendations section). Researchers and sport professionals should be cautious of taking WBV parameters direct from manufacturer's specifications on face value. This thesis has shown that theoretical and actual WBV stimuli parameters can be different. This is likely to have an impact on the design of safe and effective WBV protocols and programmes.

8.7 Recommendations

An acute bout of 60 s vertical WBV at frequencies of 30, 35, 40 and 50 Hz and a theoretical D_{PTP} of 3mm may represent too short a duration of stimuli to elicit a positive jump performance response in well-trained individuals. Further research is warranted to investigate a possible individualised response to varying WBV parameters in several participant populations; but also during different stimuli delivered by other WBV platform types. A method to investigate this could utilise EMG measures to determine which WBV frequency elicited the highest response for an individual user. The next stage of research should investigate whether the individualised WBV parameters result in an acute enhancement of performance measures.

During the process of utilising EMG measures to monitor responses to WBV of varying parameters, an effective and valid notch filter techniques should be used; in addition to the more commonly used band pass filter technique. This will ensure that artificial noise and motion artifacts are not included in the EMG signal; which would potentially overestimate the neuromuscular responses to varying WBV frequencies.

Further research is required to allude to the neuromuscular mechanisms which may explain this individualised response to WBV parameters. It seems this response is not accounted for by simple physical characteristics (such as height or BM) but may involve intrinsic muscle fibre characteristics. One such research avenue may be muscle biopsies or alternative measures to determine percentage fibre types. Additional research should be undertaken to monitor a possible individualised response to WBV frequency on neuromuscular performance within elite athletes. This research should also utilise other WBV platform types.

Researchers and sports professionals should quantify and validate the WBV stimuli by directly measuring A_R. Therefore, obtaining the true magnitude of WBV stimuli applied to the WBV platform user. This has implications in terms of athletes receiving the appropriate and accurate magnitude of WBV stimuli, whether in acute or chronic programmes.

8.8 Conclusions

The overall aim of this thesis was to investigate the characteristics of WBV stimuli and the acute neuromuscular responses to such stimuli, in an attempt to provide rationale behind safe and effective WBV protocols. By identifying a possible individualised response to WBV frequency more effective protocols can be developed.

As discussed, a method (i.e. EMG recording) to analyse the neuromuscular response to WBV stimulus should be recommended as evidence of an individualised response to both muscular activity during WBV, but also performance measures post WBV has been presented. This method may allow the optimisation of individual-specific WBV protocols and therefore subsequent optimal responses to such a training stimulus. It became clear from early work in this thesis that if utilising EMG analysis during WBV careful consideration should be taken regarding filter techniques. A notch filter is recommended to avoid over-estimation of the EMG responses due to the inclusion of noise related artifacts. The use of accelerometers placed both at platform level, to monitor the WBV output delivered by the machine attempted to account for some of the early findings presented in this thesis. The use of these at different body segments allows a potential dampening response to be monitored. This may also allow the monitoring of WBV magnitudes in relation to safety considerations, reducing the risk of potential side effects. It is vital that any WBV stimulus is verified before it is used as a training aid to ensure the application of a validated training load with the aim of enhancing performance.

Chapter 9: References

Abercromby, A. F., Amonette, W. E., Layne, C. S., Mcfarlin, B. K., Hinman, M. R. & Paloski, W. H. (2007a). Variation in neuromuscular responses during acute whole-body vibration exercise. *Med Sci Sports Exerc* **39,** 1642-1650.

Abercromby, A. F., Amonette, W. E., Layne, C. S., Mcfarlin, B. K., Hinman, M. R. & Paloski, W. H. (2007b). Vibration exposure and biodynamic responses during whole-body vibration training. *Med Sci Sports Exerc* **39,** 1794-1800.

Adamo, D. E., Martin, B. J. & Johnson, P. W. (2002). Vibration-induced muscle fatigue, a possible contribution to musculoskeletal injury. *Eur J Appl Physiol* **88,** 134-140.

Adams, J. B., Edwards, D., Serviette, D., Bedient, A. M., Huntsman, E., Jacobs, K. A., Del Rossi, G., Roos, B. A. & Signorile, J. F. (2009). Optimal frequency, displacement, duration, and recovery patterns to maximize power output following acute whole-body vibration. *J Strength Cond Res* **23,** 237-245.

Armstrong, W. J., Grinnell, D. C. & Warren, G. S. (2010). The acute effect of whole-body vibration on the vertical jump height. *J Strength Cond Res* **24,** 2835-2839.

Armstrong, W. J., Nestle, H. N., Grinnell, D. C., Cole, L. D., Van Gilder, E. L., Warren, G. S. & Capizzi, E. A. (2008). The acute effect of whole-body vibration on the hoffmann reflex. *J Strength Cond Res* **22,** 471-476.

Bagheri, J., Van Den Berg-Emons, R. J., Pel, J. J., Horemans, H. L. & Stam, H. J. (2012). Acute effects of whole-body vibration on jump force and jump rate of force development: a comparative study of different devices. *J Strength Cond Res* **26,** 691-696.

Bakhtiary, A. H., Safavi-Farokhi, Z. & Aminian-Far, A. (2007). Influence of vibration on delayed onset of muscle soreness following eccentric exercise. *Br J Sports Med* **41,** 145-148.

Bartlett, R. (2007). *Introduction to sports biomechanics : analysing human movement patterns.* Routledge, London.

Bazett-Jones, D. M., Finch, H. W. & Dugan, E. L. (2008a). Comparing the effects of various whole body vibration accelerations on counter-movement jump performance. *J Sports Sci Med* **7,** 144-150.

Bazett-Jones, D. M., Gibson, M. H. & Mcbride, J. M. (2008b). Sprint and vertical jump performances are not affected by six weeks of static hamstring stretching. *J Strength Cond Res* **22,** 25-31.

Becker, R. & Awiszus, F. (2001). Physiological alterations of maximal voluntary quadriceps activation by changes of knee joint angle. *Muscle Nerve* **24,** 667-672.

Bedient, A. M., Adams, J. B., Edwards, D. A., Serravite, D. H., Huntsman, E., Mow, S. E., Roos, B. A. & Signorile, J. F. (2009). Displacement and frequency for maximizing power output resulting from a bout of whole-body vibration. *J Strength Cond Res* **23,** 1683-1687.

Belavy, D. L., Hides, J. A., Wilson, S. J., Stanton, W., Dimeo, F. C., Rittweger, J., Felsenberg, D. & Richardson, C. A. (2008). Resistive simulated weightbearing exercise with whole body vibration reduces lumbar spine deconditioning in bed-rest. *Spine* **33,** E121-131.

Berschin, G. & Sommer, H. M. (2010). The influence of posture on transmission and absorption of vibration energy in whole body vibration exercise. *Sportverletz Sportschaden* **24,** 36-39. (Abstract accessed only).

Bevan, H. R., Cunningham, D. J., Tooley, E. P., Owen, N. J., Cook, C. J. & Kilduff, L. P. (2010). Influence of postactivation potentiation on sprinting performance in professional rugby players. *J Strength Cond Res* **24,** 701-705.

Blottner, D., Salanova, M., Puttmann, B., Schiffl, G., Felsenberg, D., Buehring, B. & Rittweger, J. (2006). Human skeletal muscle structure and function preserved by vibration muscle exercise following 55 days of bed rest. *Eur J Appl Physiol* **97,** 261-271.

Bongiovanni, L. G. & Hagbarth, K. E. (1990). Tonic vibration reflexes elicited during fatigue from maximal voluntary contractions in man. *J Physiol* **423,** 1-14.

Bongiovanni, L. G., Hagbarth, K. E. & Stjernberg, L. (1990). Prolonged muscle vibration reducing motor output in maximal voluntary contractions in man. *J Physiol* **423,** 15-26.

Bosco, C., Cardinale, M. & Tsarpela, O. (1999a). Influence of vibration on mechanical power and electromyogram activity in human arm flexor muscles. *Eur J Appl Physiol Occup Physiol* **79,** 306-311.

Bosco, C., Cardinale, M., Tsarpela, O., Colli, R., Tihanyi, J., Duvillard, S. & Viru, A. (1998). The influence of whole body vibration on jumping performance. *Bio Sport* **15,** 157-164.

Bosco, C., Colli, R., Introini, E., Cardinale, M., Tsarpela, O., Madella, A., Tihanyi, J. & Viru, A. (1999b). Adaptive responses of human skeletal muscle to vibration exposure. *Clin Physiol* **19,** 183-187.

Bosco, C., Iacovelli, M., Tsarpela, O., Cardinale, M., Bonifazi, M., Tihanyi, J., Viru, M., De Lorenzo, A. & Viru, A. (2000). Hormonal responses to whole-body vibration in men. *Eur J Appl Physiol* **81,** 449-454.

Bosco, C. & Komi, P. V. (1979). Mechanical characteristics and fiber composition of human leg extensor muscles. *Eur J Appl Physiol Occup Physiol* **41,** 275-284.

Bosco, C., Zanon, S., Rusko, H., Dal Monte, A., Bellotti, P., Latteri, F., Candeloro, N., Locatelli, E., Azzaro, E., Pozzo, R. & Et Al. (1984). The influence of extra load on the mechanical behavior of skeletal muscle. *Eur J Appl Physiol Occup Physiol* **53,** 149-154.

Bove, M., Nardone, A. & Schieppati, M. (2003). Effects of leg muscle tendon vibration on group Ia and group II reflex responses to stance perturbation in humans. *J Physiol* **550,** 617-630.

Brownstein, B., Lamb, R. E. & Mangine, R. E. (1985). Quadriceps torque and integrated electromyography. *J Orthop Sports Phys Ther* **6,** 309-314.

Brunetti, O., Filippi, G. M., Lorenzini, M., Liti, A., Panichi, R., Roscini, M., Pettorossi, V. E. & Cerulli, G. (2006). Improvement of posture stability by vibratory stimulation following anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc* **14,** 1180-1187.

Buckthorpe, M., Morris, J. & Folland, J. P. (2012). Validity of vertical jump measurement devices. *J Sports Sci* **30,** 63-69.

Bullock, N., Martin, D. T., Ross, A., Rosemond, C. D., Jordan, M. J. & Marino, F. E. (2008). Acute effect of whole-body vibration on sprint and jumping performance in elite skeleton athletes. *J Strength Cond Res* **22,** 1371-1374.

Bullock, N., Martin, D. T., Ross, A., Rosemond, C. D., Jordan, M. J. & Marino, F. E. (2009). An acute bout of whole body vibration on skeleton start and 30-m sprint performance. *Eur J Sport Sci* **9,** 35-39.

Burke, D., Hagbarth, K. E., Lofstedt, L. & Wallin, B. G. (1976a). The responses of human muscle spindle endings to vibration during isometric contraction. *J Physiol* **261,** 695-711.

Burke, D., Hagbarth, K. E., Lofstedt, L. & Wallin, B. G. (1976b). The responses of human muscle spindle endings to vibration of non-contracting muscles. *J Physiol* **261,** 673-693.

Butler, R. J., Crowell, H. P., 3rd & Davis, I. M. (2003). Lower extremity stiffness: implications for performance and injury. *Clin Biomech (Bristol, Avon)* **18,** 511-517.

Byrne, P. J., Moran, K., Rankin, P. & Kinsella, S. (2010). A comparison of methods used to identify 'optimal' drop height for early phase adaptations in depth jump training. *J Strength Cond Res* **24,** 2050-2055.

Cardinale, M. & Bosco, C. (2003). The use of vibration as an exercise intervention. *Exerc Sport Sci Rev* **31,** 3-7.

Cardinale, M. & Erskine, J. A. (2008). Vibration training in elite sport: effective training solution or just another fad? *Int J Sports Physiol Perform* **3,** 232-239.

Cardinale, M. & Lim, J. (2003a). The acute effects of two different whole body vibration frequencies on vertical jump performance. *Med Sport* **56,** 287-292.

Cardinale, M. & Lim, J. (2003b). Electromyography activity of vastus lateralis muscle during whole-body vibrations of different frequencies. *J Strength Cond Res* **17,** 621-624.

Cardinale, M. & Wakeling, J. (2005). Whole body vibration exercise: are vibrations good for you? *Br J Sports Med* **39,** 585-589.

Carson, R. G., Riek, S., Mackey, D. C., Meichenbaum, D. P., Willms, K., Forner, M. & Byblow, W. D. (2004). Excitability changes in human forearm corticospinal projections and spinal reflex pathways during rhythmic voluntary movement of the opposite limb. *J Physiol* **560,** 929-940.

Clark, D. R., Lambert, M. I. & Hunter, A. M. (2012). Muscle activation in the loaded free barbell squat: a brief review. *J Strength Cond Res* **26,** 1169-1178.

Cochrane, D. J., Legg, S. J. & Hooker, M. J. (2004). The short-term effect of whole-body vibration training on vertical jump, sprint, and agility performance. *J Strength Cond Res* **18,** 828-832.

Cochrane, D. J., Loram, I. D., Stannard, S. R. & Rittweger, J. (2009). Changes in joint angle, muscle-tendon complex length, muscle contractile tissue displacement, and modulation of EMG activity during acute whole-body vibration. *Muscle Nerve* **40,** 420-429.

Cochrane, D. J., Sartor, F., Winwood, K., Stannard, S. R., Narici, M. V. & Rittweger, J. (2008a). A comparison of the physiologic effects of acute whole-body vibration exercise in young and older people. *Arch Phys Med Rehabil* **89,** 815-821.

Cochrane, D. J. & Stannard, S. R. (2005). Acute whole body vibration training increases vertical jump and flexibility performance in elite female field hockey players. *Br J Sports Med* **39,** 860-865.

Cochrane, D. J., Stannard, S. R., Firth, E. C. & Rittweger, J. (2010). Acute whole-body vibration elicits post-activation potentiation. *Eur J Appl Physiol* **108,** 311-319.

Cochrane, D. J., Stannard, S. R., Sargeant, A. J. & Rittweger, J. (2008b). The rate of muscle temperature increase during acute whole-body vibration exercise. *Eur J Appl Physiol* **103,** 441-448.

Colson, S. S., Pensini, M., Espinosa, J., Garrandes, F. & Legros, P. (2010). Whole-body vibration training effects on the physical performance of basketball players. *J Strength Cond Res* **24,** 999-1006.

Colson, S. S., Petit, P. D., Hebreard, L., Tessaro, J. & Pensini, M. (2009). Whole body vibration does not enhance muscle activation. *Int J Sports Med* **30,** 841-844.

Cook, D. P., Mileva, K. N., James, D. C., Zaidell, L. N., Goss, V. G. & Bowtell, J. L. (2011). Triaxial modulation of the acceleration induced in the lower extremity during whole-body vibration training: a pilot study. *J Strength Cond Res* **25,** 298-308.

Cormie, P., Deane, R. S., Triplett, N. T. & Mcbride, J. M. (2006). Acute effects of whole-body vibration on muscle activity, strength, and power. *J Strength Cond Res* **20,** 257-261.

Costantino, C., Pogliacomi, F. & Soncini, G. (2006). Effect of the vibration board on the strength of ankle dorsal and plantar flexor muscles: a preliminary randomized controlled study. *Acta Biomed* **77,** 10-16.

Cram, J. & Kasman, G. (1998). *Introduction to surface electromyography*. Aspen, New York, USA.

Crewther, B., Cronin, J. & Keogh, J. (2004). Gravitational forces and whole body vibration: implications for prescription of vibratory stimulation. *Phys Ther Sport* **5,** 37-43.

Crewther, B., Kilduff, L. P., Cook, C. J., Middleton, M. K. & Bunce, P. J. (2012). Assessing the potentiating effects of isometric squats on athletic performance in a rugby trained population. In *International convention on science, education and medicine in sport*, Glasgow, UK.

Cronin, J., Oliver, M. & Mcnair, P. (2004). Muscle stiffness and injury effects of whole body vibration. *Phys Ther Sport* **5,** 68-74.

Crow, J. F., Buttifant, D., Kearny, S. G. & Hrysomallis, C. (2012). Low load exercises targeting the gluteal muscle group acutely enhance explosive power output in elite athletes. *J Strength Cond Res* **26,** 438-442.

Da Silva-Grigoletto, M. E., De Hoyo, M., Sanudo, B., Carrasco, L. & Garcia-Manso, J. M. (2011). Determining the optimal whole-body vibration dose-response relationship for muscle performance. *J Strength Cond Res* **25,** 3326-3333.

Da Silva-Grigoletto, M. E., Nunez, V. M., Vaamonde, D. M., Fernandez, J. M., Poblador, M. S., Garcia-Manso, J. M. & Lancho, J. L. (2006). Effects of different frequencies of whole body vibration on muscular performance. *Biology of Sport* **23,** 267-282.

Da Silva-Grigoletto, M. E., Vaamonde, D. M., Castillo, E., Poblador, M. S., Garcia-Manso, J. M. & Lancho, J. L. (2009). Acute and cumulative effects of different times of recovery from whole body vibration exposure on muscle performance. *J Strength Cond Res* **23,** 2073-2082.

Dabbs, N. C., Brown, L. E., Coburn, J. W., Lynn, S. K., Biagini, M. S. & Tran, T. T. (2010). Effect of whole-body vibration warm-up on bat speed in women softball players. *J Strength Cond Res* **24,** 2296-2299.

Dabbs, N. C., Munoz, C. X., Tran, T. T., Brown, L. E. & Bottaro, M. (2011). Effect of different rest intervals after whole-body vibration on vertical jump performance. *J Strength Cond Res* **25,** 662-667.

De Luca, C. J., Gilmore, L. D., Kuznetsov, M. & Roy, S. H. (2010). Filtering the surface EMG signal: Movement artifact and baseline noise contamination. *J Biomech* **43,** 1573-1579.

De Ruiter, C. J., Van Der Linden, R. M., Van Der Zijden, M. J., Hollander, A. P. & De Haan, A. (2003). Short-term effects of whole-body vibration on maximal voluntary isometric knee extensor force and rate of force rise. *Eur J Appl Physiol* **88,** 472-475.

Defreitas, J. M., Beck, T. W. & Stock, M. S. (2012). Comparison of methods for removing electromagnetic noise from electromyographic signals. *Physiol Meas* **33,** 147-158.

Delecluse, C., Roelants, M., Diels, R., Koninckx, E. & Verschueren, S. (2005). Effects of whole body vibration training on muscle strength and sprint performance in sprint-trained athletes. *Int J Sports Med* **26,** 662-668.

Delecluse, C., Roelants, M. & Verschueren, S. (2003). Strength increase after whole-body vibration compared with resistance training. *Med Sci Sports Exerc* **35,** 1033-1041.

Delsys Inc. (2008). Bagnoli EMG system user guide. Delsys Incorporated.

Devries, H. A. (1968). "Efficiency of electrical activity" as a physiological measure of the functional state of muscle tissue. *Am J Phys Med* **47,** 10-22.

Di Giminiani, R., Manno, R., Scrimaglio, R., Sementilli, G. & Tihanyi, J. (2010). Effects of individualized whole-body vibration on muscle flexibility and mechanical power. *J Sports Med Phys Fitness* **50,** 139-151.

Di Giminiani, R., Tihanyi, J., Safar, S. & Scrimaglio, R. (2009). The effects of vibration on explosive and reactive strength when applying individualized vibration frequencies. *J Sports Sci* **27,** 169-177.

Dolny, D. G. & Reyes, G. F. (2008). Whole body vibration exercise: training and benefits. *Curr Sports Med Rep* **7,** 152-157.

Dugan, E. L., Doyle, T. L., Humphries, B., Hasson, C. J. & Newton, R. U. (2004). Determining the optimal load for jump squats: a review of methods and calculations. *J Strength Cond Res* **18,** 668-674.

Ebben, W. P. & Petushek, E. J. (2010). Using the reactive strength index modified to evaluate plyometric performance. *J Strength Cond Res* **24,** 1983-1987.

Eckhardt, H., Wollny, R., Muller, H., Bartsch, P. & Friedmann-Bette, B. (2011). Enhanced myofiber recruitment during exhaustive squatting performed as whole-body vibration exercise. *J Strength Cond Res* **25,** 1120-1125.

Edwards, R. H., Hill, D. K., Jones, D. A. & Merton, P. A. (1977). Fatigue of long duration in human skeletal muscle after exercise. *J Physiol* **272,** 769-778.

Eklund, G. & Hagbarth, K. E. (1966). Normal variability of tonic vibration reflexes in man. *Exp Neurol* **16,** 80-92.

Enoka, R. M. (2002). *Neuromechanics of human movement*. Human Kinetics, Champaign.

Erskine, J., Smillie, I., Leiper, J., Ball, D. & Cardinale, M. (2007). Neuromuscular and hormonal responses to a single session of whole body vibration exercise in healthy young men. *Clin Physiol Funct Imaging* **27,** 242-248.

Fattorini, L., Ferraresi, A., Rodio, A., Azzena, G. B. & Filippi, G. M. (2006). Motor performance changes induced by muscle vibration. *Eur J Appl Physiol* **98,** 79-87.

Fauth, M. L., Petushek, E. J., Feldmann, C. R., Hsu, B. E., Garceau, L. R., Lutsch, B. N. & Ebben, W. P. (2010). Reliability of surface electromyography during maximal voluntary isometric contractions, jump landings, and cutting. *J Strength Cond Res* **24,** 1131-1137.

Field, A. (2005). *Discovering statistics using SPSS*. SAGE Publications, London.

Fitts, R. H., Riley, D. R. & Widrick, J. J. (2001). Functional and structural adaptations of skeletal muscle to microgravity. *J Exp Biol* **204,** 3201-3208.

Flanagan, E. P., Ebben, W. P. & Jensen, R. L. (2008). Reliability of the reactive strength index and time to stabilization during depth jumps. *J Strength Cond Res* **22,** 1677-1682.

Flanagan, E. P. & Harrison, A. J. (2007). Muscle dynamics differences between legs in healthy adults. *J Strength Cond Res* **21,** 67-72.

Franchignoni, F., Vercelli, S. & Ozcakar, L. (2012). Hematuria in a runner after treatment with whole body vibration: A case report. *Scand J Med Sci Sports. DOI: 10.1111/j.1600-0838.2012.01478.x.*

Fratini, A., Cesarelli, M., Bifulco, P. & Romano, M. (2009a). Relevance of motion artifact in electromyography recordings during vibration treatment. *J Electromyogr Kinesiol* **19,** 710-718.

Fratini, A., La Gatta, A., Bifulco, P., Romano, M. & Cesarelli, M. (2009b). Muscle motion and EMG activity in vibration treatment. *Med Eng Phys* **31,** 1166-1172.

Furness, T. P., Maschette, W. E., Lorenzen, C., Naughton, G. A. & Williams, M. D. (2010). Efficacy of a whole-body vibration intervention on functional performance of community-dwelling older adults. *J Altern Complement Med* **16,** 795-797.

Galileo. (2007). http://www.galileouk.co.uk/contra.html Galileo contraindications - the elite in vibration training. Retrieved 12th May 2009.

Gerdle, B., Henriksson-Larsen, K., Lorentzon, R. & Wretling, M. L. (1991). Dependence of the mean power frequency of the electromyogram on muscle force and fibre type. *Acta Physiol Scand* **142,** 457-465.

Gerodimos, V., Zafeiridis, A., Karatrantou, K., Vasilopoulou, T., Chanou, K. & Pispirikou, E. (2010). The acute effects of different whole-body vibration amplitudes and frequencies on flexibility and vertical jumping performance. *J Sci Med Sport* **13,** 438-443.

Griffin, M. (1990). *Handbook of human vibration*. London Academic Press, London.

Guggenheimer, J. D., Dickin, D. C., Reyes, G. F. & Dolny, D. G. (2009). The effects of specific preconditioning activities on acute sprint performance. *J Strength Cond Res* **23,** 1135-1139.

Gulf Coast Data Concepts. (2009). XLR8R Version 2.0 Help Documentation.

Hakkinen, K., Alen, M. & Komi, P. V. (1985). Changes in isometric force- and relaxation-time, electromyographic and muscle fibre characteristics of human skeletal muscle during strength training and detraining. *Acta Physiol Scand* **125,** 573-585.

Hamada, T., Sale, D. G., Macdougall, J. D. & Tarnopolsky, M. A. (2000). Postactivation potentiation, fiber type, and twitch contraction time in human knee extensor muscles. *J Appl Physiol* **88,** 2131-2137.

Hamano, T., Kaji, R., Diaz, A. F., Kohara, N., Takamatsu, N., Uchiyama, T., Shibasaki, H. & Kimura, J. (1993). Vibration-evoked sensory nerve action potentials derived from Pacinian corpuscles. *Electroencephalogr Clin Neurophysiol* **89,** 278-286.

Hannah, R., Minshull, C. & Folland, J. P. (2012). Whole-body vibration does not influence knee joint neuromuscular function or proprioception. *Scand J Med Sci Sports doi: 10.1111/j.1600-0838.2011.01361.x.*

Harazin, B. & Grzesik, J. (1998). The transmission of vertical whole body vibration to the body segments of stnading subjects. *Journal of Sound and Vibration* **215,** 775-787.

Harman, E. A., Rosenstein, M. T., Frykman, P. N. & Rosenstein, R. M. (1990). The effects of arms and countermovement on vertical jumping. *Med Sci Sports Exerc* **22,** 825-833.

Hazell, T. J., Jakobi, J. M. & Kenno, K. A. (2007). The effects of whole-body vibration on upper- and lower-body EMG during static and dynamic contractions. *Appl Physiol Nutr Metab* **32,** 1156-1163.

Hazell, T. J., Kenno, K. A. & Jakobi, J. M. (2010). Evaluation of muscle activity for loaded and unloaded dynamic squats during vertical whole-body vibration. *J Strength Cond Res* **24,** 1860-1865.

Hermens, H. J., Freriks, B., Disselhorst-Klug, C. & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *J Electromyogr Kinesiol* **10,** 361-374.

Hodgson, M., Dochery, D. & Robbins, D. (2005). Post-activation potentiation. *Sports Med* **35,** 585-595.

Hopkins, J. T., Fredericks, D., Guyon, P. W., Parker, S., Gage, M., Feland, J. B. & Hunter, I. (2009). Whole body vibration does not potentiate the stretch reflex. *Int J Sports Med* **30,** 124-129.

Hopkins, T., Pak, J. O., Robertshaw, A. E., Feland, J. B., Hunter, I. & Gage, M. (2008). Whole body vibration and dynamic restraint. *Int J Sports Med* **29,** 424-428.

Hopkins, W. G. (2000). Reliability from consecutive pairs of trials (Excel spreadsheet). In *A new view of statistics*, vol. 2012. International Society of Sport Science http://www.sportsci.org/resource/stats/xrely.xls.

Hopkins, W. G. (2001). Simple statistics: the spread. In *A new view of statistics*, vol. 2012. International Society of Sport Science http://sportsci.org/resource/stats/index.html.

Hough, P. A., Ross, E. Z. & Howatson, G. (2009). Effects of dynamic and static stretching on vertical jump performance and electromyographic activity. *J Strength Cond Res* **23,** 507-512.

Hunt, C. C. (1990). Mammalian muscle spindle: peripheral mechanisms. *Physiol Rev* **70,** 643-663.

Iso. (1997). *ISO 2631-1 Mechanical Vibration and Shock - Evaluation of Human Exposure to Whole Body Vibration, Part 1, General Requirements*. International Organization for Standardization, Geneva, Switzerland.

Issurin, V. B. (2005). Vibrations and their applications in sport. A review. *J Sports Med Phys Fitness* **45,** 324-336.

Issurin, V. B., Liebermann, D. G. & Tenenbaum, G. (1994). Effect of vibratory stimulation training on maximal force and flexibility. *J Sports Sci* **12,** 561-566.

Issurin, V. B. & Tenenbaum, G. (1999). Acute and residual effects of vibratory stimulation on explosive strength in elite and amateur athletes. *J Sports Sci* **17,** 177-182.

Jackson, S. W. & Turner, D. L. (2003). Prolonged muscle vibration reduces maximal voluntary knee extension performance in both the ipsilateral and the contralateral limb in man. *Eur J Appl Physiol* **88,** 380-386.

Jacobs, P. L. & Burns, P. (2009). Acute enhancement of lower-extremity dynamic strength and flexibility with whole-body vibration. *J Strength Cond Res* **23,** 51-57.

Jordan, M. J., Norris, S. R., Smith, D. J. & Herzog, W. (2005). Vibration training: an overview of the area, training consequences, and future considerations. *J Strength Cond Res* **19,** 459-466.

Jordon, M., Norris, S., Smith, D. & Herzog, W. (2010). Acute effects of whole-body vibration on peak isometric torque, muscle twitch torque and voluntary muscle activation of the knee extensors. *Scand J Med Sci Sports* **20,** 535-540.

Kamen, G. & Gabriel, D. (2010). *Essentials of electromyography*. Human Kinetics, Leeds, UK.

Karlsson, S. & Gerdle, B. (2001). Mean frequency and signal amplitude of the surface EMG of the quadriceps muscles increase with increasing torque--a study using the continuous wavelet transform. *J Electromyogr Kinesiol* **11,** 131-140.

Kasai, T., Kawanishi, M. & Yahagi, S. (1992). The effects of wrist muscle vibration on human voluntary elbow flexion-extension movements. *Exp Brain Res* **90,** 217-220.

Katz, R., Penicaud, A. & Rossi, A. (1991). Reciprocal Ia inhibition between elbow flexors and extensors in the human. *J Physiol* **437,** 269-286.

Kelly, S. B., Alvar, B. A., Black, L. E., Dodd, D. J., Carothers, K. F. & Brown, L. E. (2010). The effect of warm-up with whole-body vibration vs. cycle ergometry on isokinetic dynamometry. *J Strength Cond Res* **24,** 3140-3143.

Kemertzis, M. A., Lythgo, N. D., Morgan, D. L. & Galea, M. P. (2008). Ankle Flexors Produce Peak Torque at Longer Muscle Lengths after Whole-Body Vibration. *Med Sci Sports Exerc* **40,** 1977 - 1983.

Khadra, M. H., Pickard, R. S., Charlton, M., Powell, P. H. & Neal, D. E. (2000). A prospective analysis of 1,930 patients with hematuria to evaluate current diagnostic practice. *J Urol* **163,** 524-527.

Kiiski, J., Heinonen, A., Jarvinen, T. L., Kannus, P. & Sievanen, H. (2008). Transmission of vertical whole body vibration to the human body. *J Bone Miner Res* **23,** 1318-1325.

Kilduff, L. P., Bevan, H. R., Kingsley, M. I., Owen, N. J., Bennett, M. A., Bunce, P. J., Hore, A. M., Maw, J. R. & Cunningham, D. J. (2007). Postactivation potentiation in professional rugby players: optimal recovery. *J Strength Cond Res* **21,** 1134-1138.

Kilduff, L. P., Owen, N., Bevan, H., Bennett, M., Kingsley, M. I. & Cunningham, D. (2008). Influence of recovery time on post-activation potentiation in professional rugby players. *J Sports Sci* **26,** 795-802.

Kinser, A. M., Ramsey, M. W., O'bryant, H. S., Ayres, C. A., Sands, W. A. & Stone, M. H. (2008). Vibration and stretching effects on flexibility and explosive strength in young gymnasts. *Med Sci Sports Exerc* **40,** 133-140.

Kipp, K., Johnson, S. T. & Hoffman, M. A. (2012). Spectral properties of h-reflex recordings after an acute bout of whole-body vibration. *J Strength Cond Res* **26,** 1915-1919.

Kitazaki, S. & Griffin, M. J. (1995). A data correction method for surface measurement of vibration on the human body. *J Biomech* **28,** 885-890.

Knikou, M. (2008). The H-reflex as a probe: pathways and pitfalls. *J Neurosci Methods* **171,** 1-12.

Komi, P. V. & Vitasalo, J. H. (1976). Signal characteristics of EMG at different levels of muscle tension. *Acta Physiol Scand* **96,** 267-276.

Kovacs, I., Tihanyi, J., Devita, P., Racz, L., Barrier, J. & Hortobagyi, T. (1999). Foot placement modifies kinematics and kinetics during drop jumping. *Med Sci Sports Exerc* **31,** 708-716.

Kreighbaum, E. & Barthels, K. (1996). *Biomechanics : a qualitative approach for studying human movement* Allyn and Bacon, London.

Laffaye, G., Bardy, B. & Taiar, R. (2006). Upper-limb motion and drop jump: effect of expertise. *J Sports Med Phys Fitness* **46,** 238-247.

Laffaye, G. & Choukou, M. A. (2010). Gender bias in the effect of dropping height on jumping performance in volleyball players. *J Strength Cond Res* **24,** 2143-2148.

Lambertz, D., Perot, C., Kaspranski, R. & Goubel, F. (2001). Effects of long-term spaceflight on mechanical properties of muscles in humans. *J Appl Physiol* **90,** 179-188.

Lamont, H. S., Cramer, J. T., Bemben, D. A., Shehab, R. L., Anderson, M. A. & Bemben, M. G. (2009). Effects of a 6-week periodized squat training program with or without whole-body vibration on jump height and power output following acute vibration exposure. *J Strength Cond Res* **23,** 2317-2325.

Lamont, H. S., Cramer, J. T., Bemben, D. A., Shehab, R. L., Anderson, M. A. & Bemben, M. G. (2010a). The acute effect of whole-body low-frequency vibration on countermovement vertical jump performance in college-aged men. *J Strength Cond Res* **24,** 3433-3442.

Lamont, H. S., Cramer, J. T., Bemben, D. A., Shehab, R. L., Anderson, M. A. & Bemben, M. G. (2010b). Effects of adding whole body vibration to squat training on isometric force/time characteristics. *J Strength Cond Res* **24,** 171-183.

Lau, R. W., Liao, L. R., Yu, F., Teo, T., Chung, R. C. & Pang, M. Y. (2011). The effects of whole body vibration therapy on bone mineral density and leg muscle strength in older adults: a systematic review and meta-analysis. *Clin Rehabil* **25,** 975-988.

Lees, A., Vanrenterghem, J. & De Clercq, D. (2004). Understanding how an arm swing enhances performance in the vertical jump. *J Biomech* **37,** 1929-1940.

Linthorne, N. P. (2001). Analysis of standing vertical jumps using a force platform. *American Journal of Physics* **69,** 1198-1204.

Lorenzen, C., Maschette, W., Koh, M. & Wilson, C. (2009). Inconsistent use of terminology in whole body vibration exercise research. *J Sci Med Sport* **12,** 676-678.

Lovell, R., Midgley, A., Barrett, S., Carter, D. & Small, K. (2012). Effects of different half-time strategies on second half soccer-specific speed, power and dynamic strength. *Scand J Med Sci Sports DOI 10.111/j.1600-0838.2011.01353.x*.

Luo, J., Clarke, M., Mcnamara, B. & Moran, K. (2009). Influence of resistance load on neuromuscular response to vibration training. *J Strength Cond Res* **23,** 420-426.

Luo, J., Mcnamara, B. & Moran, K. (2005a). The use of vibration training to enhance muscle strength and power. *Sports Med* **35,** 23-41.

Luo, J., Mcnamara, B. & Moran, K. (2008). Effect of vibration training on neuromuscular output with ballistic knee extensions. *J Sports Sci* **26,** 1365-1373.

Luo, J., Mcnamara, B. P. & Moran, K. (2005b). A portable vibrator for muscle performance enhancement by means of direct muscle tendon stimulation. *Med Eng Phys* **27,** 513-522.

Lythgo, N., Eser, P., De Groot, P. & Galea, M. (2009). Whole-body vibration dosage alters leg blood flow. *Clin Physiol Funct Imaging* **29,** 53-59.

Marin, P. J., Bunker, D., Rhea, M. R. & Ayllon, F. N. (2009). Neuromuscular activity during whole-body vibration of different amplitudes and footwear conditions: implications for prescription of vibratory stimulation. *J Strength Cond Res* **23,** 2311-2316.

Marin, P. J., Herrero, A. J., Sainz, N., Rhea, M. R. & Garcia-Lopez, D. (2010). Effects of different magnitudes of whole-body vibration on arm muscular performance. *J Strength Cond Res* **24,** 2506-2511.

Marin, P. J. & Rhea, M. R. (2010a). Effects of vibration training on muscle power: a meta-analysis. *J Strength Cond Res* **24,** 871-878.

Marin, P. J. & Rhea, M. R. (2010b). Effects of vibration training on muscle strength: a meta-analysis. *J Strength Cond Res* **24,** 548-556.

Marin, P. J., Santos-Lozano, A., Santin-Medeiros, F., Delecluse, C. & Garatachea, N. (2011). A comparison of training intensity between whole-body vibration and conventional squat exercise. *J Electromyogr Kinesiol* **21,** 616-621.

Matsumoto, Y. & Griffin, M. (1998). Dynamic response of the standing human body exposed to vertical vibration: influence of posture and vibration magnitude. *J Sound & Vib* **212,** 85-107.

Matthews, P. B. (1966). The reflex excitation of the soleus muscle of the decerebrate cat caused by vibbration applied to its tendon. *J Physiol* **184,** 450-472.

Matthews, P. B. (1991). The human stretch reflex and the motor cortex. *Trends Neurosci* **14,** 87-91.

Mcbride, J. M., Cormie, P. & Deane, R. (2006). Isometric squat force output and muscle activity in stable and unstable conditions. *J Strength Cond Res* **20,** 915-918.

Mcbride, J. M., Nuzzo, J. L., Dayne, A. M., Israetel, M. A., Nieman, D. C. & Triplett, N. T. (2010). Effect of an acute bout of whole body vibration exercise on muscle force output and motor neuron excitability. *J Strength Cond Res* **24,** 184-189.

Mcbride, J. M., Porcari, J. P. & Scheunke, M. D. (2004). Effect of vibration during fatiguing resistance exercise on subsequent muscle activity during maximal voluntary isometric contractions. *J Strength Cond Res* **18,** 777-781.

Mccaw, S. T. & Melrose, D. R. (1999). Stance width and bar load effects on leg muscle activity during the parallel squat. *Med Sci Sports Exerc* **31,** 428-436.

Mcclung, M. & Collins, D. (2007). "Because I know it will!": placebo effects of an ergogenic aid on athletic performance. *J Sport Exerc Psychol* **29,** 382-394.

Mcdonagh, M. J. & Davies, C. T. (1984). Adaptive response of mammalian skeletal muscle to exercise with high loads. *Eur J Appl Physiol Occup Physiol* **52,** 139-155.

Mclellan, C. P., Lovell, D. I. & Gass, G. C. (2011). The role of rate of force development on vertical jump performance. *J Strength Cond Res* **25,** 379-385.

Mcneal, J. R. & Sands, W. A. (2006). Stretching for performance enhancement. *Curr Sports Med Rep* **5,** 141-146.

Mellor, R. & Hodges, P. W. (2006). Effect of knee joint angle on motor unit synchronization. *J Orthop Res* **24,** 1420-1426.

Melnyk, M., Kofler, B., Faist, M., Hodapp, M. & Gollhofer, A. (2008). Effect of a whole-body vibration session on knee stability. *Int J Sports Med* **29,** 839-844.

Merletti, R. & Parker, P. (2004). *Electromyography: Physiology, engineering and non-invasive applications*. Wiley-Interscience, New York.

Mester, J., Kleinoder, H. & Yue, Z. (2006). Vibration training: benefits and risks. *J Biomech* **39,** 1056-1065.

Mewett, D. T., Reynolds, K. J. & Nazeran, H. (2004). Reducing power line interference in digitised electromyogram recordings by spectrum interpolation. *Med Biol Eng Comput* **42,** 524-531.

Mileva, K. N., Bowtell, J. L. & Kossev, A. R. (2009). Effects of low-frequency whole-body vibration on motor-evoked potentials in healthy men. *Exp Physiol* **94,** 103-116.

Mileva, K. N., Naleem, A. A., Biswas, S. K., Marwood, S. & Bowtell, J. L. (2006). Acute effects of a vibration-like stimulus during knee extension exercise. *Med Sci Sports Exerc* **38,** 1317-1328.

Mischi, M. & Cardinale, M. (2009). The effects of a 28-Hz vibration on arm muscle activity during isometric exercise. *Med Sci Sports Exerc* **41,** 645-652.

Mischi, M., Rabotti, C. & Cardinale, M. (2010). Electromyographic assessment of muscle fatigue during isometric vibration training at varying frequencies. *Conf Proc IEEE Eng Med Biol Soc* **2010,** 2338-2341.

Misiaszek, J. E. (2003). The H-reflex as a tool in neurophysiology: its limitations and uses in understanding nervous system function. *Muscle Nerve* **28,** 144-160.

Moezy, A., Olyaei, G., Hadian, M., Razi, M. & Faghihzadeh, S. (2008). A comparative study of whole body vibration training and conventional training on knee proprioception and postural stability after anterior cruciate ligament reconstruction. *Br J Sports Med* **42,** 373-378.

Monteleone, G., De Lorenzo, A., Sgroi, M., De Angelis, S. & Di Renzo, L. (2007). Contraindications for whole body vibration training: a case of nephrolitiasis. *J Sports Med Phys Fitness* **47,** 443-445.

Moran, K., Mcnamara, B. & Luo, J. (2007). Effect of vibration training in maximal effort (70% 1RM) dynamic bicep curls. *Med Sci Sports Exerc* **39,** 526-533.

Moras, G., Rodriguez-Jimenez, S., Tous-Fajardo, J., Ranz, D. & Mujika, I. (2010). A vibratory bar for upper body: feasibility and acute effects on EMGrms activity. *J Strength Cond Res* **24,** 2132-2142.

Moritani, T. & Devries, H. A. (1979). Neural factors versus hypertrophy in the time course of muscle strength gain. *Am J Phys Med* **58,** 115-130.

Nigg, B. M. & Wakeling, J. M. (2001). Impact forces and muscle tuning: a new paradigm. *Exerc Sport Sci Rev* **29,** 37-41.

Nordin, M. & Hagbarth, K. E. (1996). Effects of preceding movements and contractions on the tonic vibration reflex of human finger extensor muscles. *Acta Physiol Scand* **156,** 435-440.

Nordlund, M. M. & Thorstensson, A. (2007). Strength training effects of whole-body vibration? *Scand J Med Sci Sports* **17,** 12-17.

Onishi, H., Yagi, R., Akasaka, K., Momose, K., Ihashi, K. & Handa, Y. (2000). Relationship between EMG signals and force in human vastus lateralis muscle using multiple bipolar wire electrodes. *J Electromyogr Kinesiol* **10,** 59-67.

Payton, C. & Bartlett, R. (2008). *Biomechanical evaluation of movement in sport and exercise. The BASES guidelines*. Routledge, Oxon, UK.

Pel, J. J., Bagheri, J., Van Dam, L. M., Van Den Berg-Emons, H. J., Horemans, H. L., Stam, H. J. & Van Der Steen, J. (2009). Platform accelerations of three different whole-body vibration devices and the transmission of vertical vibrations to the lower limbs. *Med Eng Phys* **31,** 937-944.

Pellegrini, M. J., Lythgo, N. D., Morgan, D. L. & Galea, M. P. (2010). Voluntary activation of the ankle plantar flexors following whole-body vibration. *Eur J Appl Physiol* **108,** 927-934.

Petit, P. D., Pensini, M., Tessaro, J., Desnuelle, C., Legros, P. & Colson, S. S. (2010). Optimal whole-body vibration settings for muscle strength and power enhancement in human knee extensors. *J Electromyogr Kinesiol* **20,** 1186-1195.

Pincivero, D. M., Salfetnikov, Y., Campy, R. M. & Coelho, A. J. (2004). Angle- and gender-specific quadriceps femoris muscle recruitment and knee extensor torque. *J Biomech* **37,** 1689-1697.

Pocock, G. & Richards, C. (1999). *Human physiology: The basis of medicine*. Oxford university press, Oxford.

Pollock, R. D., Woledge, R. C., Martin, F. C. & Newham, D. J. (2012). Effects of whole body vibration on motor unit recruitment and threshold. *J Appl Physiol* **112,** 388-395.

Pollock, R. D., Woledge, R. C., Mills, K. R., Martin, F. C. & Newham, D. J. (2010). Muscle activity and acceleration during whole body vibration: effect of frequency and amplitude. *Clin Biomech (Bristol, Avon)* **25,** 840-846.

Powerplate. (2013). Powerplate: A true 3D workout.

Preatoni, E., Colombo, A., Verga, M., Galvani, C., Faina, M., Rodano, R. & Cardinale, M. (2012). The Effects of Whole Body Vibration in Isolation or Combined with Strength Training in Female Athletes. *J Strength Cond Res* **26,** 2495-2506.

Prisby, R. D., Lafage-Proust, M. H., Malaval, L., Belli, A. & Vico, L. (2008). Effects of whole body vibration on the skeleton and other organ systems in man and animal models: what we know and what we need to know. *Ageing Res Rev* **7,** 319-329.

Rauch, F., Sievanen, H., Boonen, S., Cardinale, M., Degens, H., Felsenberg, D., Roth, J., Schoenau, E., Verschueren, S. & Rittweger, J. (2010). Reporting whole-body vibration intervention studies: recommendations of the International Society of Musculoskeletal and Neuronal Interactions. *J Musculoskelet Neuronal Interact* **10,** 193-198.

Read, M. M. & Cisar, C. (2001). The influence of varied rest interval lengths on depth jump performance. *J Strength Cond Res* **15,** 279-283.

Rehn, B., Lidstrom, J., Skoglund, J. & Lindstrom, B. (2007). Effects on leg muscular performance from whole-body vibration exercise: a systematic review. *Scand J Med Sci Sports* **17,** 2-11.

Reiser, R. F., Rocheford, E. C. & Armstrong, C. J. (2006). Building a better understanding of basic mechanical principles through analysis of the vertical jump. *Strength and Conditioning Journal* **28,** 70-80.

Rhea, M. R. & Kenn, J. G. (2009). The effect of acute applications of whole-body vibration on the iTonic platform on subsequent lower-body power output during the back squat. *J Strength Cond Res* **23,** 58-61.

Rittweger, J. (2010). Vibration as an exercise modality: how it may work, and what its potential might be. *Eur J Appl Physiol* **108,** 877-904.

Rittweger, J., Beller, G. & Felsenberg, D. (2000). Acute physiological effects of exhaustive whole-body vibration exercise in man. *Clin Physiol* **20,** 134-142.

Rittweger, J., Mutschelknauss, M. & Felsenberg, D. (2003). Acute changes in neuromuscular excitability after exhaustive whole body vibration exercise as compared to exhaustion by squatting exercise. *Clin Physiol Funct Imaging* **23,** 81-86.

Ritzmann, R., Gollhofer, A. & Kramer, A. (2012). The influence of vibration type, frequency, body position and additional load on the neuromuscular activity during whole body vibration. *Eur J Appl Physiol DOI 10.1007/s00421-012-2402-0.*

Ritzmann, R., Kramer, A., Gruber, M., Gollhofer, A. & Taube, W. (2010). EMG activity during whole body vibration: motion artifacts or stretch reflexes? *Eur J Appl Physiol* **110,** 143-151.

Robbins, D. W. (2005). Postactivation potentiation and its practical applicability: a brief review. *J Strength Cond Res* **19,** 453-458.

Roelants, M., Verschueren, S. M., Delecluse, C., Levin, O. & Stijnen, V. (2006). Whole-body-vibration-induced increase in leg muscle activity during different squat exercises. *J Strength Cond Res* **20,** 124-129.

Ronnestad, B. R. (2004). Comparing the performance-enhancing effects of squats on a vibration platform with conventional squats in recreationally resistance-trained men. *J Strength Cond Res* **18,** 839-845.

Ronnestad, B. R. (2009). Acute effects of various whole-body vibration frequencies on lower-body power in trained and untrained subjects. *J Strength Cond Res* **23,** 1309-1315.

Ronnestad, B. R. & Ellefsen, S. (2011). The effects of adding different whole-body vibration frequencies to preconditioning exercise on subsequent sprint performance. *J Strength Cond Res* **25,** 3306-3310.

Ronnestad, B. R., Holden, G., Samnoy, L. E. & Paulsen, G. (2012). Acute effect of whole-body vibration on power, one-repetition maximum, and muscle activation in power lifters. *J Strength Cond Res* **26,** 531-539.

Ross, E. Z., Middleton, N., Shave, R., George, K. & Nowicky, A. (2007). Corticomotor excitability contributes to neuromuscular fatigue following marathon running in man. *Exp Physiol* **92,** 417-426.

Rubin, C., Pope, M., Fritton, J. C., Magnusson, M., Hansson, T. & Mcleod, K. (2003). Transmissibility of 15-hertz to 35-hertz vibrations to the human hip and lumbar spine: determining the physiologic feasibility of delivering low-level anabolic mechanical stimuli to skeletal regions at greatest risk of fracture because of osteoporosis. *Spine (Phila Pa 1976)* **28,** 2621-2627.

Rutherford, O. M. & Jones, D. A. (1986). The role of learning and coordination in strength training. *Eur J Appl Physiol Occup Physiol* **55,** 100-105.

Sale, D. (2004). Postactivation potentiation: role in performance. *Br J Sports Med* **38,** 386-387.

Sale, D. G. (1988). Neural adaptation to resistance training. *Med Sci Sports Exerc* **20,** S135-145.

Sale, D. G. (2002). Postactivation potentiation: role in human performance. *Exerc Sport Sci Rev* **30,** 138-143.

Sands, W. A., Mcneal, J. R., Stone, M. H., Russell, E. M. & Jemni, M. (2006). Flexibility enhancement with vibration: Acute and long-term. *Med Sci Sports Exerc* **38,** 720-725.

Sanudo, B., Feria, A., Carrasco, L., De Hoyo, M., Santos, R. & Gamboa, H. (2012). Gender differences in knee stability in response to whole-body vibration. *J Strength Cond Res* **26,** 2156-2165.

Savelberg, H. H., Keizer, H. A. & Meijer, K. (2007). Whole-body vibration induced adaptation in knee extensors; consequences of initial strength, vibration frequency, and joint angle. *J Strength Cond Res* **21,** 589-593.

Sebik, O., Karacan, I., Cidem, M. & Turker, K. S. (2013). Rectification of SEMG as a tool to demonstrate synchronous motor unit activity during vibration. *J Electromyogr Kinesiol* **23,** 275-284.

Simic, L., Sarabon, N. & Markovic, G. (2012). Does pre-exercise static stretching inhibit maximal muscular performance? A meta-analytical review. *Scand J Med Sci Sports DOI: 10.1111/j.1600-0838.2012.01444.x.*

Sitja Rabert, M., Rigau Comas, D., Fort Vanmeerhaeghe, A., Santoyo Medina, C., Roque, I. F. M., Romero-Rodriguez, D. & Bonfill Cosp, X. (2012). Whole-body vibration training for patients with neurodegenerative disease. *Cochrane Database Syst Rev* **2,** CD009097.

Stewart, J. A., Cochrane, D. J. & Morton, R. H. (2009). Differential effects of whole body vibration durations on knee extensor strength. *J Sci Med Sport* **12,** 50-53.

Sweeney, H. L., Bowman, B. F. & Stull, J. T. (1993). Myosin light chain phosphorylation in vertebrate striated muscle: regulation and function. *Am J Physiol* **264,** C1085-1095.

Thompson, C. W. & Floyd, R. T. (1998). *Manual of structural kinesiology*. McGraw-Hill, Singapore.

Till, K. A. & Cooke, C. (2009). The effects of postactivation potentiation on sprint and jump performance of male academy soccer players. *J Strength Cond Res* **23,** 1960-1967.

Torvinen, S., Kannu, P., Sievanen, H., Jarvinen, T. A., Pasanen, M., Kontulainen, S., Jarvinen, T. L., Jarvinen, M., Oja, P. & Vuori, I. (2002a). Effect of a vibration exposure on muscular performance and body balance. Randomized cross-over study. *Clin Physiol Funct Imaging* **22,** 145-152.

Torvinen, S., Kannus, P., Sievanen, H., Jarvinen, T. A., Pasanen, M., Kontulainen, S., Jarvinen, T. L., Jarvinen, M., Oja, P. & Vuori, I. (2002b). Effect of four-month vertical whole body vibration on performance and balance. *Med Sci Sports Exerc* **34,** 1523-1528.

Torvinen, S., Sievanen, H., Jarvinen, T. A., Pasanen, M., Kontulainen, S. & Kannus, P. (2002c). Effect of 4-min vertical whole body vibration on muscle performance and body balance: a randomized cross-over study. *Int J Sports Med* **23,** 374-379.

Trans, T., Aaboe, J., Henriksen, M., Christensen, R., Bliddal, H. & Lund, H. (2009). Effect of whole body vibration exercise on muscle strength and proprioception in females with knee osteoarthritis. *Knee* **16,** 256-261.

Trappe, S., Costill, D., Gallagher, P., Creer, A., Peters, J. R., Evans, H., Riley, D. A. & Fitts, R. H. (2009). Exercise in space: human skeletal muscle after 6 months aboard the International Space Station. *J Appl Physiol* **106,** 1159-1168.

Turner, A. P., Sanderson, M. F. & Attwood, L. A. (2011). The acute effect of different frequencies of whole-body vibration on countermovement jump performance. *J Strength Cond Res* **25,** 1592-1597.

Wakeling, J. M. & Nigg, B. M. (2001). Modification of soft tissue vibrations in the leg by muscular activity. *J Appl Physiol* **90,** 412-420.

Wakeling, J. M., Nigg, B. M. & Rozitis, A. I. (2002). Muscle activity damps the soft tissue resonance that occurs in response to pulsed and continuous vibrations. *J Appl Physiol* **93,** 1093-1103.

Wallace, B. J., Kernozek, T. W., White, J. M., Kline, D. E., Wright, G. A., Peng, H. T. & Huang, C. F. (2010). Quantification of vertical ground reaction forces of popular bilateral plyometric exercises. *J Strength Cond Res* **24,** 207-212.

Wallmann, H. W., Mercer, J. A. & Landers, M. R. (2008). Surface electromyographic assessment of the effect of dynamic activity and dynamic activity with static stretching of the gastrocnemius on vertical jump performance. *J Strength Cond Res* **22,** 787-793.

Wallmann, H. W., Mercer, J. A. & Mcwhorter, J. W. (2005). Surface electromyographic assessment of the effect of static stretching of the gastrocnemius on vertical jump performance. *J Strength Cond Res* **19,** 684-688.

Walsh, M., Arampatzis, A., Schade, F. & Bruggemann, G. P. (2004). The effect of drop jump starting height and contact time on power, work performed, and moment of force. *J Strength Cond Res* **18,** 561-566.

Wilcock, I. M., Whatman, C., Harris, N. & Keogh, J. W. (2009). Vibration training: could it enhance the strength, power, or speed of athletes? *J Strength Cond Res* **23,** 593-603.

Winter, D. (1990). *Biomechanics and motor control of human movement*. Wiley-Interscience, New York.

Young, W. (1995). Laboratory strength assessment of athletes. *New Studies in Athletics* **10,** 88 - 96.

Young, W., Cormack, S. & Crichton, M. (2011). Which jump variables should be used to assess explosive leg muscle function? *Int J Sports Physiol Perform* **6,** 51-57.

Zelena, J. (1994). *Nerves and mechanoreceptors. The role of innervation in the development and maintainence of mammalian mechanoreceptor*. Chapman & Hall, London.

Zhi, G., Ryder, J. W., Huang, J., Ding, P., Chen, Y., Zhao, Y., Kamm, K. E. & Stull, J. T. (2005). Myosin light chain kinase and myosin phosphorylation effect frequency-dependent potentiation of skeletal muscle contraction. *Proc Natl Acad Sci U S A* **102,** 17519-17524.

Chapter 10: Appendices

Chapter 3 Appendix: Ethical approval letter

Research & Knowledge Exchange
MORAY HOUSE SCHOOL *of* EDUCATION

The University of Edinburgh
Old Moray House
Holyrood Road
Edinburgh EH8 8AQ

Direct Dial 0131 651 6388
Switchboard 0131 650 1000
Email rke-education@ed.ac.uk
http://www.ed.ac.uk

Mark Sanderson
SPEHS
St Leonards Land

24th August 2011

Dear Mark

Investigating the influence of loading on acceleration and electromyography during the whole body vibration at different frequencies

The School of Education Ethics Sub-Committee has now considered your request for ethical approval for the studies detailed in the your application.

This is to confirm that the Sub-Committee is happy to approve the application and that the research meets the School Ethics Level 2 criterion. This is defined as“covering novel procedures or the use of atypical participant groups – usually projects in which ethical issues might require more detailed consideration but were unlikely to prove problematic”.

Ethical approval is granted subject to amendment of the participants’ information to clarify that not taking part in, or withdrawing from, the research will not affect the student’s status on the High Performance Programme.

A standard condition of this ethical approval is that you are required to notify the Committee, of any significant proposed deviation from the original protocol. The Committee also needs to be notified if there are any unexpected results or events once the research is underway that raise questions about the safety of the research.

Yours sincerely

S Cunningham

pp Dr K McCulloch
Convener, School Ethics Sub-Committee

The University of Edinburgh is a charitable body, registered in Scotland, with registration number SC005336

Chapter 3 Appendix

PARTICPANT INFORMATION SHEET – Study part 1

We would like to invite you to take part in our research study. Before you decide we would like you to understand why the research is being done and what it would involve.

What is the purpose of the study?

To investigate the effects of whole body vibration (WBV) on athletic performance. Whole body vibration is a new type of training. It is thought to cause your muscles to contract while you stand on the platform. This is thought to affect explosive athletic performance such as how high you can jump. This study will investigate the effects of different frequencies of whole body vibration on your jumping ability.

Why have I been invited?

Because you are a healthy, active 16 to 35 year old male and have a history of resistance training. During the study you will not be asked to change your normal physical activity.

What will happen to me if I take part?

You will be asked to attend the sport science laboratory on five different testing days. The first visit will give you the chance to familiarise yourself with the jump tests and with the WBV platform. The first part of the study involves 5 visits. During these you will be tested on your jumping performance before and after WBV.

Each test day will involve the following:

- A 5 minute warm up on the bike and plyometric warm up.
- A maximal contraction of knee extension and then knee flexion.
- 6 maximal jump tests on a force platform. This involves jumping up from the floor but also off a small height landing on the floor and jumping up in the air again as quickly as possible.
- Holding a squat position on the platform for 60 seconds of WBV with a rest of 60 seconds.
- 6 maximal drop jump tests as before.

During the visit you will be asked to wear electromyography (EMG) pads which are placed on the skin to measure electrical activity of the muscle. EMG recordings will be taken during the jump tests and while you stand on the WBV platform. The EMG pads will not be used to electrically stimulate your muscles.

The platform will be set at a different WBV frequency each visit however, the frequency will be concealed from you. Each visit will last approximately 45 minutes to 1 hour, and will be separated by at least 24 hours.

Are there any side effects?

WBV has the potential for side effects if used incorrectly. You will be shown the correct way to stand to minimise the risk of these side effects. In past studies a very small number of participants experienced temporary redness and itching of the legs lasting a few minutes after WBV. WBV may not improve the jumping performance of everyone.

Chapter 3 Appendix

There are a number of contra-indications that are used to exclude participants who may be at risk of developing more serious side effects to WBV.

Do I have to take part?

It is up to you decide to join the study. If you agree to take part, we will then ask you to sign a consent form. You are free to withdraw at any time, without giving a reason, and all information you have provided will be destroyed. All ethical and legal practices will be followed. The data collected will be used in a PhD thesis and potentially in a scientific journal publication. All the data will be anonymous.

M.F.Sanderson@sms.ed.ac.uk Tel: 0131 650 9791

Chapter 3 Appendix

PARTICIPANT CONSENT FORM – Study part 1

Title of Project:

The acute effects of individualised whole body vibration (WBV) frequencies on electromyography (EMG) responses and neuromuscular performance during explosive jump tests.

Name of Researcher:

Mark Sanderson

Please state yes or no if you have experienced any of the following:

Recent fractures or recent injuries that stopped you training	Yes / No
Wearing a pacemaker	Yes / No
Hernia	Yes / No
Recent surgery or recent wounds	Yes / No
Deep vein thrombosis	Yes / No
Cardiovascular disease	Yes / No
Hip or knee joint replacements	Yes / No
Spinal medical conditions such as discopathy or spondylosis	Yes / No
Epilepsy	Yes / No
Known neurological conditions	Yes / No
Joint or bone implants such as metal pins or plates	Yes / No
Diabetes	Yes / No
Tumours	Yes / No
A history of gallstones or kidney stones	Yes / No

I confirm that I have read and understand the information sheet for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.

I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason.

I agree to take part in the above study and know of no reason why I should not take part.

Name of Participant……………………………………………………………………………..

Signature………………………………………………………………………………………

Date ………………………………….

Name of Person taking consent……………………………………………………………….

Signature………………………………………………………………………………………

Date …………………………………

Chapter 3 Appendix

PARTICPANT INFORMATION SHEET – Study part 2

We would like to invite you to take part in the second part of a research study. Before you decide we would like you to understand why the research is being done and what it would involve.

What is the purpose of the study – part 2?

To investigate the effects of using an individualised frequency of whole body vibration (WBV) on your jumping ability. WBV is thought to cause your muscles to contract while you stand on the platform. This is thought to increase explosive athletic performance such as how high you can jump.

However, it seems that this effect may be very individual and some people will increase their jump performance at certain frequencies. While other people will need different WBV frequencies to increase their jump performance.

Why have I been invited?

Because you have completed the first part of the study. From those five visits your individualised WBV frequency has been determined. The individualised WBV frequency was the frequency which gave you the largest increase in your jump performance and largest increase in muscle activity. This frequency is different for each person.

We can use this individualised frequency to test how much it improves your jumping ability. Previous studies have shown the WBV using an individualised frequency for that person results in a better jump performance after WBV than a set frequency given to everybody.

What will happen to me if I take part?

You will be asked to attend the sport science laboratory twice. You will complete the same maximal knee extension and flexion tests and the same jumping tests that you did during the first part of the study before using the WBV platform. You will also complete the same number of WBV repetitions.

One of the visits we will use your individualised frequency. The other visit we will not use the individualised frequency but select another frequency. After each WBV session you will be then complete the same jumping tests. Each visit will last approximately 45 minutes, and will be separated by at least 2 days.

Are there any side effects?

WBV has the potential for side effects if used incorrectly. You will be shown the correct way to stand to minimise the risk of these side effects. In past studies a very small number of participants experienced temporary redness and itching of the legs lasting a few minutes after WBV. WBV may not improve the jumping performance of everyone.

There are a number of contra-indications that are used to exclude participants who may be at risk of developing more serious side effects to WBV. You will not be experience any frequencies that you have not already experienced during the first part of the study.

Chapter 3 Appendix

Do I have to take part?

It is up to you decide to join the study. If you agree to take part, we will then ask you to sign a consent form. You are free to withdraw at any time, without giving a reason, and all information you have provided will be destroyed. All ethical and legal practices will be followed. The data collected will be used in a PhD thesis and potentially in a scientific journal publication. All the data will anonymous.

M.F.Sanderson@sms.ed.ac.uk Tel: 0131 650 9791

PARTICIPANT CONSENT FORM – Study part 2

Title of Project:

The acute effects of individualised whole body vibration (WBV) frequencies on electromyography (EMG) responses and neuromuscular performance during explosive jump tests.

Name of Researcher:

Mark Sanderson

Please state yes or no if you have experienced any of the following:

Recent fractures or recent injuries that stopped you training	Yes / No
Wearing a pacemaker	Yes / No
Hernia	Yes / No
Recent surgery or recent wounds	Yes / No
Deep vein thrombosis	Yes / No
Cardiovascular disease	Yes / No
Hip or knee joint replacements	Yes / No
Spinal medical conditions such as discopathy or spondylosis	Yes / No
Epilepsy	Yes / No
Known neurological conditions	Yes / No
Joint or bone implants such as metal pins or plates	Yes / No
Diabetes	Yes / No
Tumours	Yes / No
A history of gallstones or kidney stones	Yes / No
Uncomfortable symptoms following WBV sessions	Yes / No

I confirm that I have read and understand the information sheet for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.

I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason.

I agree to take part in the above study and know of no reason why I should not take part.

Name of Participant …………………………………………………………………………..

Signature ………………………………………………………………………………………

Date ………………………………….

Name of Person taking consent ……………………………………………………………….

Signature ………………………………………………………………………………………

Date …………………………………

Chapter 3 Appendix

PARTICPANT INFORMATION SHEET – Whole body vibration

We would like to invite you to take part in our research study. Before you decide we would like you to understand why the research is being done and what it would involve.

What is the purpose of the study?

To investigate the effects of adding weight while using a whole body vibration (WBV) platform. Whole body vibration is a new type of training. It is thought to cause your muscles to contract while you stand on the platform. However, it is unclear what happens to the vibration when load is applied onto the platform, for example when people of different weight stand on it. The actual "dose" of vibration that a person receives may be different to that stated by the machine. For a new training method it is important to know what amount of training you are receiving to correctly prescribe a correct training stimulus.

Why have I been invited?

Because you are a healthy, active 18 to 35 year old male and have a history of resistance training. During the study you will not be asked to change your normal physical activity.

What will happen to me if I take part?

You will be asked to attend the sport science laboratory once for approximately 45 minutes to one hour.

The test day will involve the following:

- Wearing clothing you can exercise in, shorts are required.
- A 5 minute warm up on the bike plus some squatting warm up exercises.
- Recording your height and weight.
- 12 reps of 20 seconds 90º squats while on the WBV platform with 2 minutes recovery between.
- The 12 reps will consist of 3 different loads at 4 different WBV frequencies.
- The 3 loads will be your body mass plus added weights using a weighted vest to make up a total of 80, 90 and 100kg.
- The 4 WBV frequencies will be 0, 30, 40 and 50Hz, these will be concealed from you but you will be told at the end of the testing.

During the testing an accelerometer will be placed on the platform and two placed on your skin at thigh and lower leg. These accelerometers measure the g-forces while the WBV platform is on.

You will also be asked to wear electromyography (EMG) pads which are placed on the skin to measure electrical activity of the muscle. EMG recordings will be taken while you stand on the WBV platform. The EMG pads will not be used to electrically stimulate your muscles.

Are there any side effects?

WBV has the potential for side effects if used incorrectly. You will be shown the correct way to stand to minimise the risk of these side effects. In past studies a very small number of participants experienced temporary redness and itching of the legs lasting a few minutes after WBV. You are free to withdraw at any time.

Chapter 3 Appendix

Both the EMG sensors and accelerometer sensors pose little risk of side effects as both are placed on the skin and are easily removed. There are a number of contra-indications that we will use to exclude participants who may be at risk of developing more serious side effects to WBV.

Do I have to take part?

It is up to you decide to join the study. There is some evidence to suggest that WBV can result in short term benefits to explosive strength and power and increase your muscle activity. However; not all studies have found this and any benefits after using WBV for a short time are temporary. If you agree to take part, we will then ask you to sign a consent form. You are free to withdraw at any time, without giving a reason, and all information you have provided will be destroyed. All ethical and legal practices will be followed. The data collected will be used in a PhD thesis, potentially in a scientific journal publication and a summary of findings which will be provided to you. However; your name will not appear in any of these.

Mark.Sanderson@sms.ed.ac.uk Tel: 07738 305333

PARTICIPANT CONSENT FORM – Whole body vibration

Title of Project:

Investigating the influence of loading on acceleration and electromyography during whole body vibration at different frequencies

Name of Researcher:

Mark Sanderson

Please state yes or no if you have experienced any of the following:

Recent fractures or recent injuries that stopped you training	Yes / No
Wearing a pacemaker	Yes / No
Hernia	Yes / No
Recent surgery or recent wounds	Yes / No
Deep vein thrombosis	Yes / No
Cardiovascular disease	Yes / No
Hip or knee joint replacements	Yes / No
Spinal medical conditions such as discopathy or spondylosis	Yes / No
Epilepsy	Yes / No
Known neurological conditions	Yes / No
Joint or bone implants such as metal pins or plates	Yes / No
Diabetes	Yes / No
Tumours	Yes / No
A history of gallstones or kidney stones	Yes / No

I confirm that I have read and understand the information sheet for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.

I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason.

I agree to take part in the above study and know of no reason why I should not take part.

Name of Participant …………………………………………………………………………..

Signature ………………………………………………………………………………………

Date ………………………………….

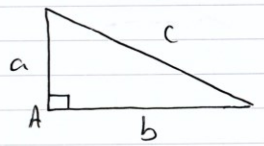
Name of Person taking consent ………………………………………………………………

Signature ………………………………………………………………………………………

Date …………………………………

Chapter 7 Appendix

Resultant acceleration calculations, page 1.



$a^2 + b^2 = c^2$

eg if 2 forces act on point A

1 force = a = 10

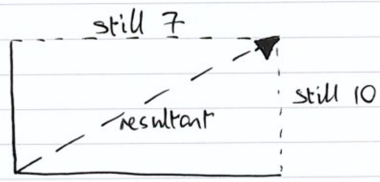
1 force = b = 7

what is resultant force

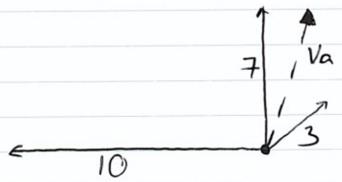
$a^2 + b^2 = c^2$

$10^2 + 7^2 = 149$

$\sqrt{149} = 12.20$



so for 3 forces on an object, calculate the resultant

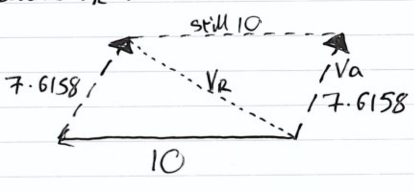


i) Calculate resultant from 2 forces of 7 & 3 (Va):

$7^2 + 3^2 = 58$

$\sqrt{58} = 7.6158$

ii) Calculate V_R :



V_R = resultant

$= 7.6158^2 + 10^2$

$= 158.000$

$\sqrt{158} = 12.570$

$V_R = 12.570$

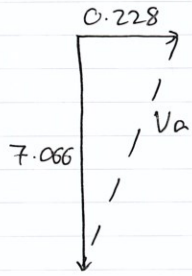
continue…..

Chapter 7 Appendix

Resultant acceleration calculations, page 2. Adapted from Kreighbaum & Barthels (1996).

Eg: WBV accelerometer data:

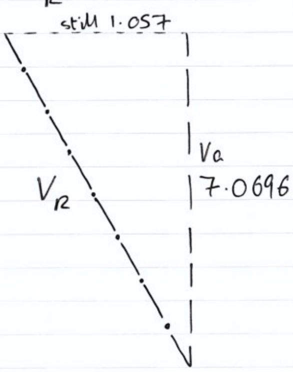
i) For Va:



$0.228^2 + 7.066^2$
$= 49.98034$
$\sqrt{49.98034} = 7.0696$

$V_a = 7.0696$

ii) For V_R



$1.057^2 + 7.0696^2$
$= 51.09649$
$\sqrt{51.09649} = 7.148$

$V_R = 7.148\,g$

So for this 20s bout of WBV the participant received 7.148g

7.148g = 70.12 ms^{-2} acceleration